02-20-04

PTO/SB/21 (08-00)
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FEB 19 2004 JC102 PATENT & TRADEMARK OFFICE

TRANSMITTAL FORM

(to be used for all correspondence after initial filing)

CUSTOMER NO. 23446

Total Number of Pages in This Submission	

Application Number	10/705,419
Filing Date	11/10/2003
First Named Inventor	Torp et al.
Group Art Unit	Not Yet Assigned
Examiner Name	Not Yet Assigned
Attorney Docket Number	15-DS-000523 DIV 2-1 12483US05

ENCLOSURES (check all that apply)

- ☒ **Fee Transmittal Form**
  - ☒ **Fee Attached (by check)** No. 14499
- ☐ Amendment/Reply
  - ☐ After Final
  - ☐ Affidavits/declaration(s)
- ☐ Extension of Time Request
- ☐ Express Abandonment Request
- ☐ Information Disclosure Statement
  - ☐ PTO 1449/08A with references
- ☐ Certified Copy of Priority Document(s)
- ☒ **Response to Notice of Incomplete Nonprovisional Application**
  - ☐ Response to Missing Parts under 37 CFR 1.52 or 1.53
- ☐ Assignment Papers (for an Application)
- ☒ **Drawing(s) (11 sheets)**
- ☐ Licensing-related Papers
- ☒ **Petition Under 37 CFR 1.182 and MPEP 201.06(c)**
- ☐ Petition to Convert to a Provisional Application
- ☒ **Copy of Declaration and Power from prior U.S. Patent Application No. 10/054,033, and as filed with Divisional Application No. 10/705,419 on November 10, 2003, which claims priority of the 10/054,033 application).**
- ☐ Terminal Disclaimer
- ☐ Request for Refund
- ☐ CD Number of CD(s) _____
- ☐ After Allowance Communication to Group
- ☐ Appeal Communication to Board of Appeals and Interferences
- ☐ Appeal Communication to Group *(Appeal Notice, Brief, Reply Brief)*
- ☐ Proprietary Information
- ☐ Status Letter
- ☒ **Other Enclosure(s) (please identify below):**

Remarks	**3 PAGES OF PETITION UNDER 37 CFR 1.182 AND MPEP 201.06(C); TAB 1=2 PAGES of Notice of Incomplete Nonprovisional Application; TAB 2=11 Sheets of Drawings; TAB 3=1 PAGE of Utility Patent Application Transmittal Form; TAB 4=6 PAGES of Copy of Declaration; TAB 5=61 PAGES of a copy of U.S. Patent Application No. 10/054,033 as filed with the U.S. Patent Office on 01/22/2002.**

SIGNATURE OF APPLICANT, ATTORNEY, OR AGENT

Firm or Individual Name	McAndrews Held & Malloy, Ltd.		
Name (Print/type)	Peter J. McAndrews	Registration No. (Attorney/Agent)	38,547
Signature		Date: 02/19/2004	

EXPRESS MAIL DEPOSIT

"Express Mail" mailing label number : **EV 327683417 US**
Date of Deposit **February 19, 2004.**

OIPE
FEB 19 2004
PATENT & TRADEMARK OFFICE

PTO/SB/17 (11-00)Approved for use through 10/31/2002. OMB 0651-0032
U.S. Patent and Trademark Office; U.S. DEPARTMENT OF COMMERCE
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FEE TRANSMITTAL for FY 2004

CUSTOMER NO. 23446

Patent Fees are subject to annual revision.

TOTAL AMOUNT OF PAYMENT ($)130.00

Complete if Known	
Application Number	10/705,419
Filing Date	11/10/2003
First Named Inventor	Torp et al.
Examiner Name	Not Yet Assigned
Group Art Unit	Not Yet Assigned
Attorney Docket No.	15-DS-000523 DIV 2-1 12483US05

METHOD OF PAYMENT

1. ☐ The Commissioner is hereby authorized to charge indicated fees and credit any overpayments to:

Deposit Account Number: 13-0017
Deposit Account Name: McAndrews Held & Malloy

☒ Charge Any Additional Fee Required Under 37 CFR 1.16 and 1.17
☐ Applicant claims small entity status. See 37 CFR 1.27

2. ☒ **Payment Enclosed:**
☒ Check ☐ Credit Card ☐ Money Order ☐ Other

FEE CALCULATION

1. **BASIC FILING FEE**

Large Entity Fee Code	Large Entity Fee ($)	Small Entity Fee Code	Small Entity Fee ($)	Fee Description	Fee Paid
1001	770	2001	385	Utility filing Fee	
1002	340	2002	170	Design filing Fee	
1003	530	2003	265	Plant filing fee	
1004	770	2004	385	Reissue filing fee	
1005	160	2005	80	Provisional filing fee	

SUBTOTAL (1) ($)0

2. EXTRA CLAIM FEES

		Extra Claims		Fee from below		Fee Paid
Total Claims	[ ] - 20** =	[ ]	x	[ ]	=	
Independent Claims	[ ] - 3** =	[ ]	x	[ ]	=	
Multiple Dependent				[ ]	=	

Large Entity Fee Code	Large Entity Fee ($)	Small Entity Fee Code	Small Entity Fee ($)	Fee Description
1202	18	2202	9	Claims in excess of 20
1201	86	2201	43	Independent claims in excess of 3
1203	290	2203	145	Multiple dependent claim, if not paid
1204	86	2204	43	**Reissue independent claims over original patent
1205	18	2205	9	**Reissue claims in excess of 20 and over original patent

SUBTOTAL (2) ($)0

**or number previously paid, if greater; For Reissues, see above

FEE CALCULATION (continued)

3. **ADDITIONAL FEES**

Large Entity Fee Code	Large Entity Fee ($)	Small Entity Fee Code	Small Entity Fee ($)	Fee Description	Fee Paid
1051	130	2051	65	Surcharge - late filing fee or oath	
1052	50	2052	25	Surcharge - late provisional filing fee or cover sheet	
1053	130	1053	130	Non-English specification	
1812	2,520	1812	2,520	For filing a request for *ex parte* reexamination	
1804	920*	1804	920*	Requesting publication of SIR prior to Examiner action	
1805	1,840*	1805	1,840*	Requesting publication of SIR after Examiner action	
1251	110	2251	55	Extension for reply within first month	
1252	420	2252	210	Extension for reply within second month	
1253	950	2253	475	Extension for reply within third month	
1254	1,480	2254	740	Extension for reply within fourth month	
1255	2,010	2255	1,005	Extension for reply within fifth month	
1401	330	2401	165	Notice of Appeal	
1402	330	2402	165	Filing a brief in support of an appeal	
1403	290	2403	145	Request for oral hearing	
1451	1,510	1451	1510	Petition to institute a public use proceeding	
1452	110	2452	55	Petition to revive - unavoidable	
1453	1,330	2453	665	Petition to revive - unintentional	
1501	1,330	2501	665	Utility issue fee (or reissue)	
1502	480	2502	240	Design issue fee	
1503	640	2503	320	Plant issue fee	
1460	130	1460	130	Petitions to the Commissioner	130.00
1807	50	1807	50	Processing fee under 37 CFR 1.17(q)	
1806	180	1806	180	Submission of Information Disclosure Stmt	
8021	40	8021	40	Recording each patent assignment per property (times number of properties)	
1809	770	2809	385	Filing a submission after final rejection (37 CFR § 1.129(a))	
1810	770	2810	385	For each additional invention to be examined (37 CFR 1.129(b))	
1801	770	2801	385	Request for Continued Examination (RCE)	
1802	900	1802	900	Request for expedited examination of a design application	
Other fee (specify)					

*Reduced by Basic Filing Fee Paid

SUBTOTAL (3) ($)130.00

SUBMITTED BY Complete (*if applicable*)

Name (Print/Type)	Peter J. McAndrews	Registration No. (Attorney or Agent)	38,547	Telephone	312-775-8000
Signature				Date	02/19/2004

WARNING: Information on this form may become public. Credit card information should not be included on this form. Provide credit card information and authorization on PTO-2038.

OIPE
FEB 19 2004
JC102
PATENT & TRADEMARK OFFICE

Attorney Docket No.: 15-DS-000523 Div 2-1

12483US05

IN THE UNITED STATES PATENT AND TRADEMARK OFFICE

In the Application of:	
Torp et al.	Examiner: Not Yet Assigned
Serial No.: 10/705,419	Group Art Unit: Not Yet Assigned
Filed: 11/10/2003	Conf. No.: 5172
For: METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING	

PETITION UNDER 37 CFR 1.182

AND

MPEP § 201.06(c)

Mail Stop Petitions
Commissioner for Patents
P.O. Box 1450
Alexandria, VA 22313-1450

Dear Sirs:

This Petition, filed pursuant to 37 CFR 1.182 and MPEP § 201.06(c), is in response to a Notice of Incomplete Nonprovisional Application, dated January 5, 2004 (attached hereto at Tab 1).

The Notice states that a filing date has not been accorded to the above-identified application papers because the application was deposited without drawings. For the following

02/25/2004 WASFAW1 00000054 10705419
01 FC:1460 130.00 OP

reasons, applicant respectfully requests that (1) the above-identified application be accorded its original filing date of November 10, 2003, and (2) the enclosed 11 sheets of drawings (attached hereto at Tab 2) be deemed part of the above-identified application as if such drawings were filed with the application.

1) The Utility Patent Application Transmittal form (attached hereto at Tab 3) submitted with the above-identified patent application states that:

> "This application is a Divisional Application of U.S. Patent Application No. 10/054,033, filed January 22, 2002, now pending; which is a Divisional of U.S. Patent Application No. 09/432,061, filed November 2, 1999, issued as U.S. Patent No. 6,352,507, which is based upon U.S. Provisional Application No. 60/150,265, filed August 23, 1999."

2) The Utility Patent Application Transmittal form (Tab 3) submitted with the above-identified patent application also states that:

> "For CONTINUATION OR DIVISIONAL APPS only: The entire disclosure of the prior application, from which an oath or declaration is supplied under Box 5b, is considered a part of the disclosure of the accompanying continuation or divisional application and is hereby incorporated by reference. The incorporation can only be relied upon when a portion has been inadvertently omitted from the submitted application parts."

3) Applicant submitted with the above-identified patent application a copy of the Declaration (attached hereto as Tab 4) from the prior U.S. Patent Application No. 10/054,033, filed January 22, 2002, and checked Box 5b on the Utility Patent Application Transmittal form (Tab 3).

4) U.S. Patent Application No. 10/054,033, which is attached hereto as Tab 5, included drawings. Applicant intended that the drawings of U.S. Patent Application No. 10/054,033 be filed as part of the above-identified application. Such drawings, however, were inadvertently omitted from the submitted application parts for the above-identified application.

Therefore, applicant respectfully requests that the above-identified application be (1) given an effective filing date of November 10, 2003, the original deposit date of the above-identified application and (2) amended to include the enclosed drawings (11 sheets attached hereto at Tab 2), which are an identical copy of the drawings of U.S. Patent Application No. 10/054,033.

Enclosed is our Check No. 14499 in the amount of $130.00 to cover the Petition Filing Fees Under 1.17(h), namely, Petitions to the Commissioner, unless otherwise specified. The Commissioner is authorized to charge any necessary fees or credit any overpayment to the Deposit Account of McAndrews, Held & Malloy, Account No. 13-0017.

Should there be any questions regarding the instant Petition, please call Peter J. McAndrews at (312) 775-8269.

Respectfully submitted,

McANDREWS, HELD & MALLOY, LTD.

Dated: February 19, 2004

Peter J. McAndrews
Reg. No. 38,547

McAndrews, Held & Malloy, Ltd.
500 West Madison Street
34th Floor
Chicago, IL 60661
Telephone: (312) 775-8000
Facsimile: (312) 775-8100

UNITED STATES PATENT AND TRADEMARK OFFICE

UNITED STATES DEPARTMENT OF COMMERCE
United States Patent and Trademark Office
Address: COMMISSIONER FOR PATENTS
P.O. Box 1450
Alexandria, Virginia 22313-1450
www.uspto.gov

APPLICATION NUMBER	FILING OR 371 (c) DATE	FIRST NAMED APPLICANT	ATTORNEY DOCKET NUMBER
10/705,419	11/10/2003	Hans Torp	15-DS-000523 DIV2-1 (1248

CONFIRMATION NO. 5172

23446
MCANDREWS HELD & MALLOY, LTD
500 WEST MADISON STREET
SUITE 3400
CHICAGO, IL 60661

FORMALITIES LETTER

OC000000011617981

Date Mailed: 01/05/2004

NOTICE OF INCOMPLETE NONPROVISIONAL APPLICATION

FILED UNDER 37 CFR 1.53(b)

A filing date has NOT been accorded to the above-identified application papers for the reason(s) indicated below.

All of the items noted below **and a newly executed oath or declaration covering the items must** be submitted within **TWO MONTHS** of the date of this Notice, unless otherwise indicated, or proceedings on the application will be terminated *(37 CFR 1.53(e))*. Replies should be mailed to: Mail Stop Missing Parts, Commissioner for Patents, P.O. Box 1450, Alexandria VA 22313-1450.

The filing date will be the date of receipt of all items required below, unless otherwise indicated. Any assertions that the item(s) required below were submitted, or are not necessary for a filing date, must be by way of petition directed to the attention of the Office of Petitions accompanied by the $130.00 petition fee *(37 CFR 1.17(h))*. If the petition states that the application is entitled to a filing date, a request for a refund of the petition fee may be included in the petition. Petitions should be mailed to: Mail Stop Petitions, Commissioner for Patents, P.O. Box 1450, Alexandria VA 22313-1450.

- The application was deposited without drawings. 35 U.S.C. 113 (first sentence) requires a drawing "where necessary for the understanding of the subject matter sought to be patented." *Applicant should reconsider whether the drawings are necessary under 35 U.S.C. 113 (first sentence).*

Replies should be mailed to: Mail Stop Missing Parts
Commissioner for Patents
P.O. Box 1450
Alexandria VA 22313-1450

A copy of this notice MUST be returned with the reply.

Customer Service Center

Initial Patent Examination Division (703) 308-1202
PART 2 - COPY TO BE RETURNED WITH RESPONSE

PATENT & TRADEMARK OFFICE
FEB 19 2004
JC102

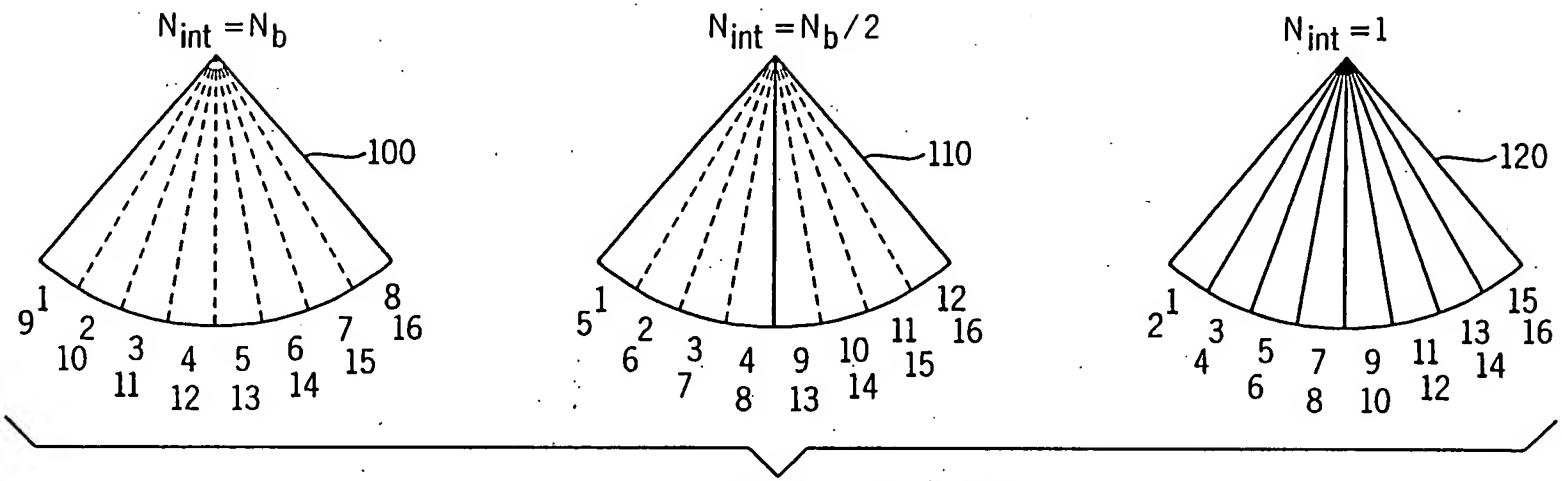


FIG. 1 (PRIOR ART)

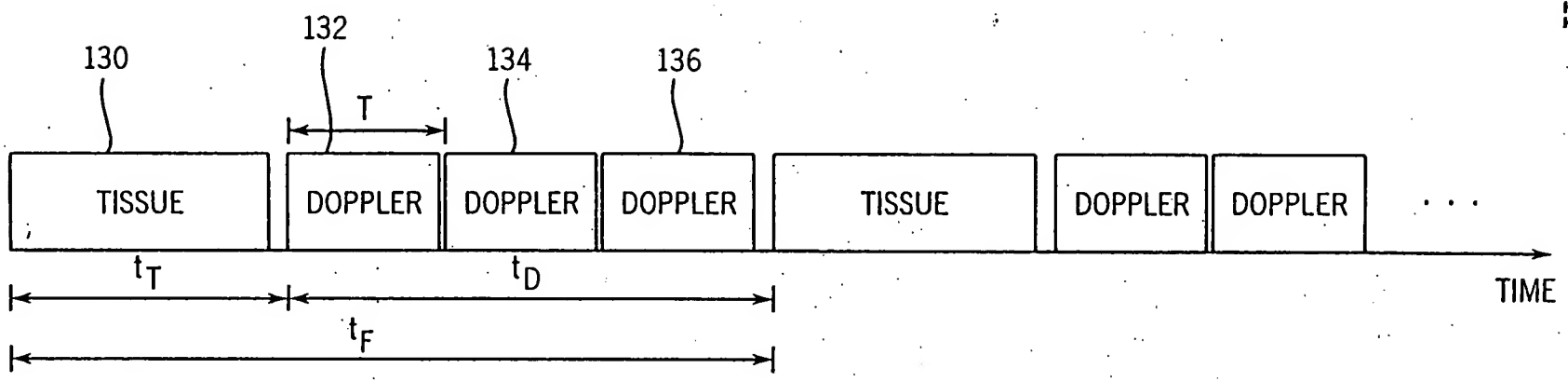


FIG. 2 (PRIOR ART)

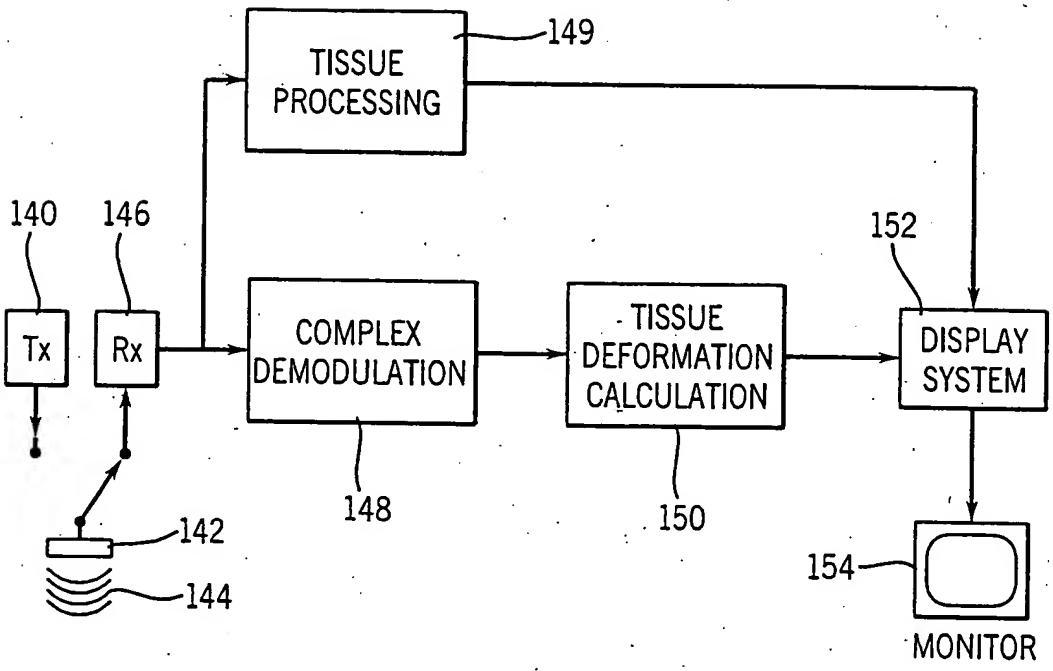

TISSUE PROCESSING
149
140
146
152
Tx
Rx
COMPLEX DEMODULATION
TISSUE DEFORMATION CALCULATION
DISPLAY SYSTEM
148
150
142
144
154
MONITOR

FIG. 3

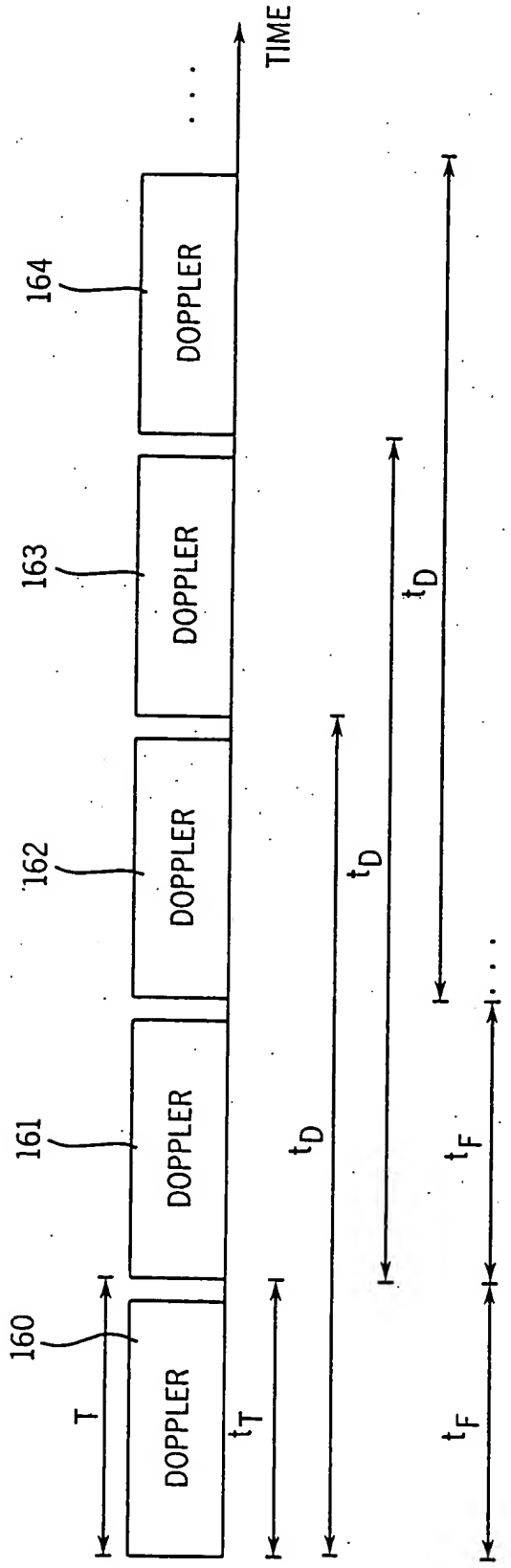


FIG. 4

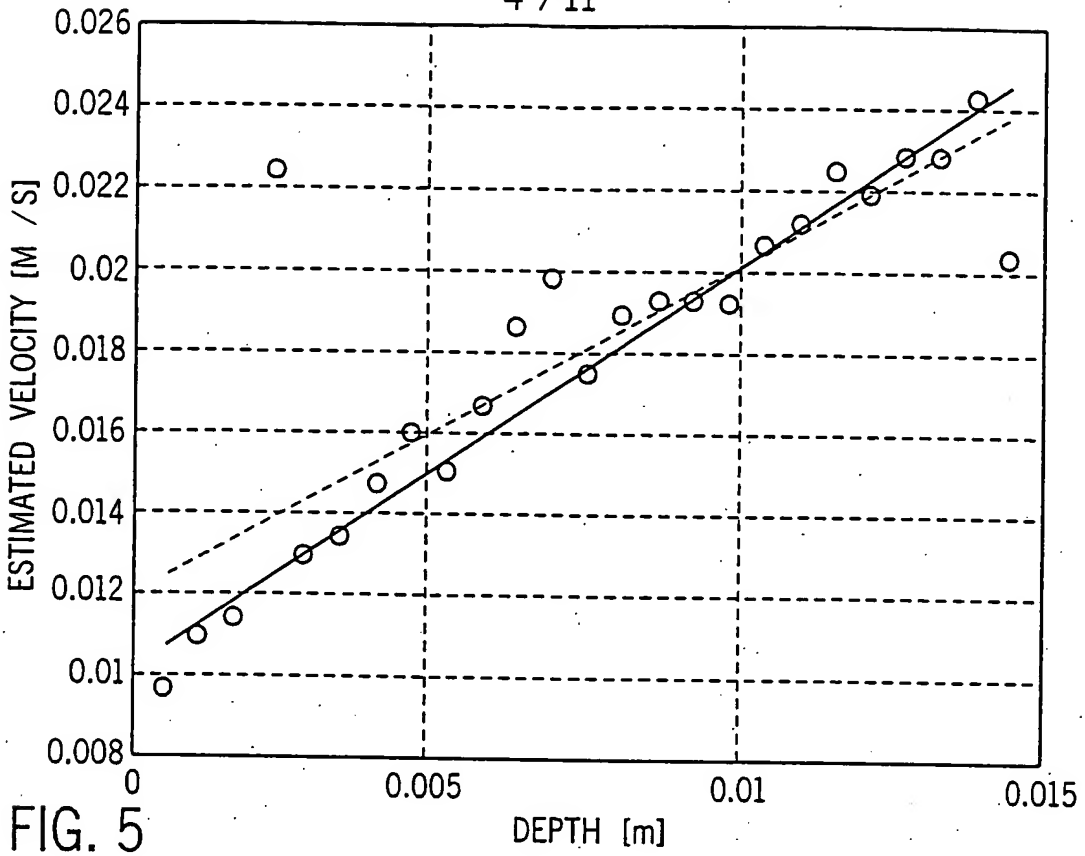

0.026
0.024
0.022
0.02
0.018
0.016
0.014
0.012
0.01
0.008
ESTIMATED VELOCITY [M /S]
0
0.005
0.01
0.015
DEPTH [m]

FIG. 5

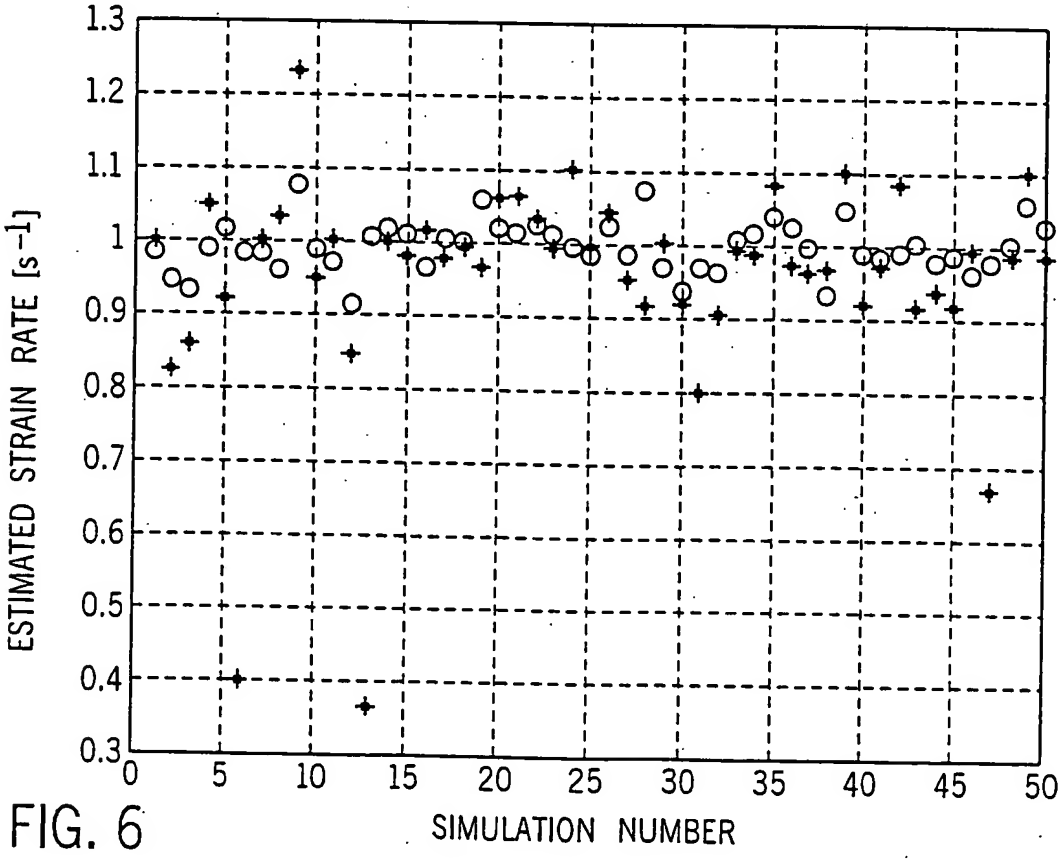

1.3
1.2
1.1
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
ESTIMATED STRAIN RATE [s^{-1}]
0
5
10
15
20
25
30
35
40
45
50
SIMULATION NUMBER

FIG. 6

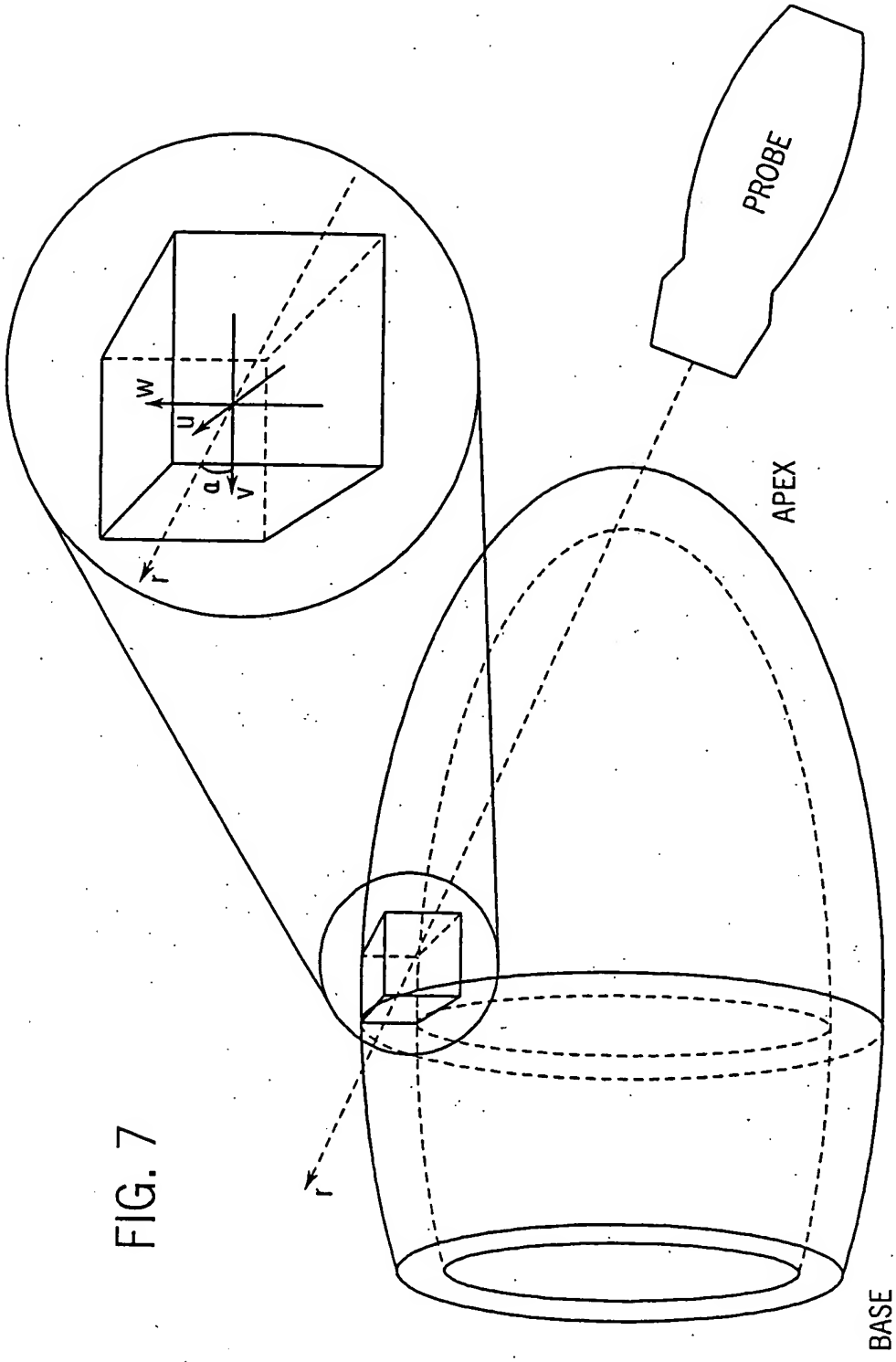


FIG. 7

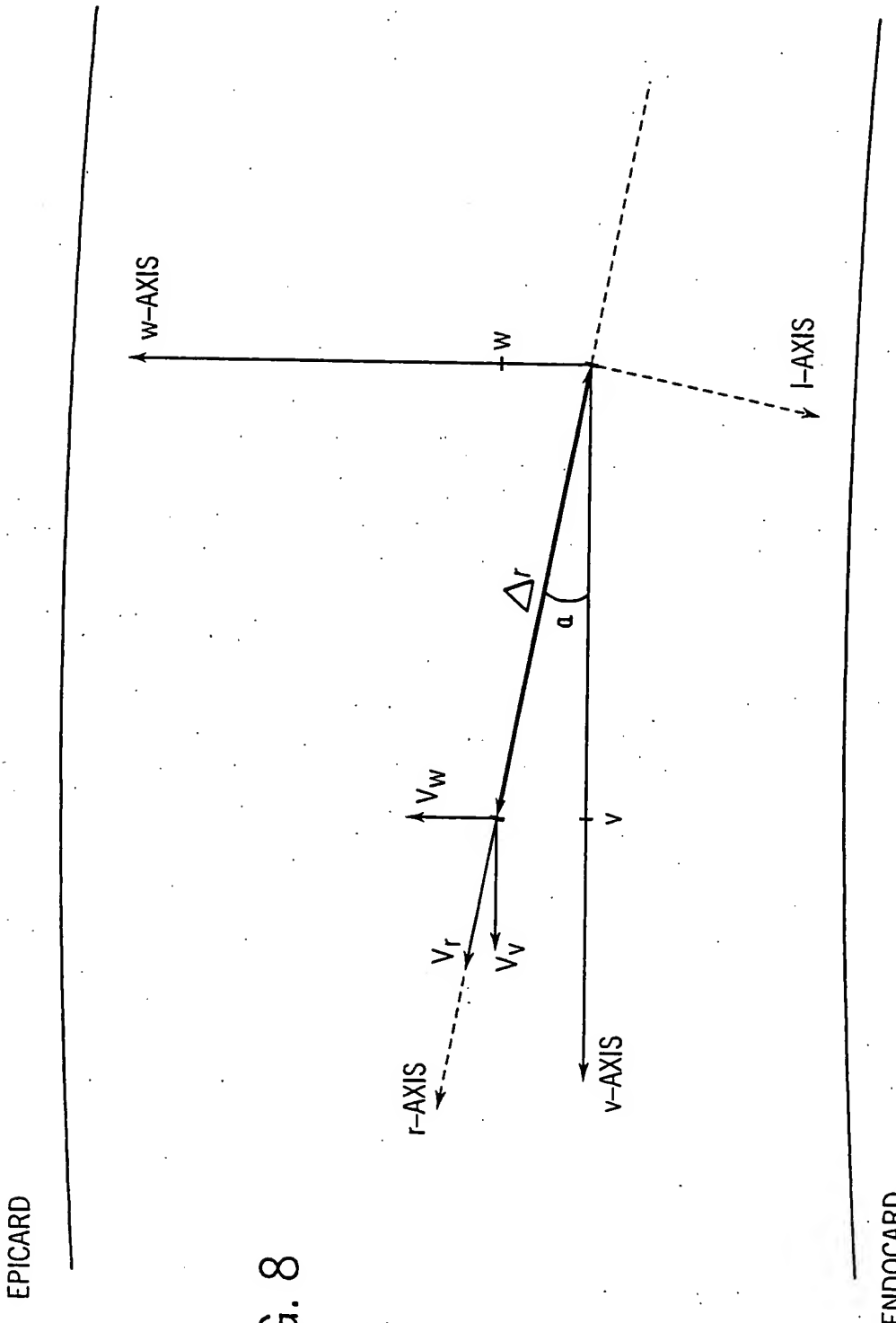


FIG. 8

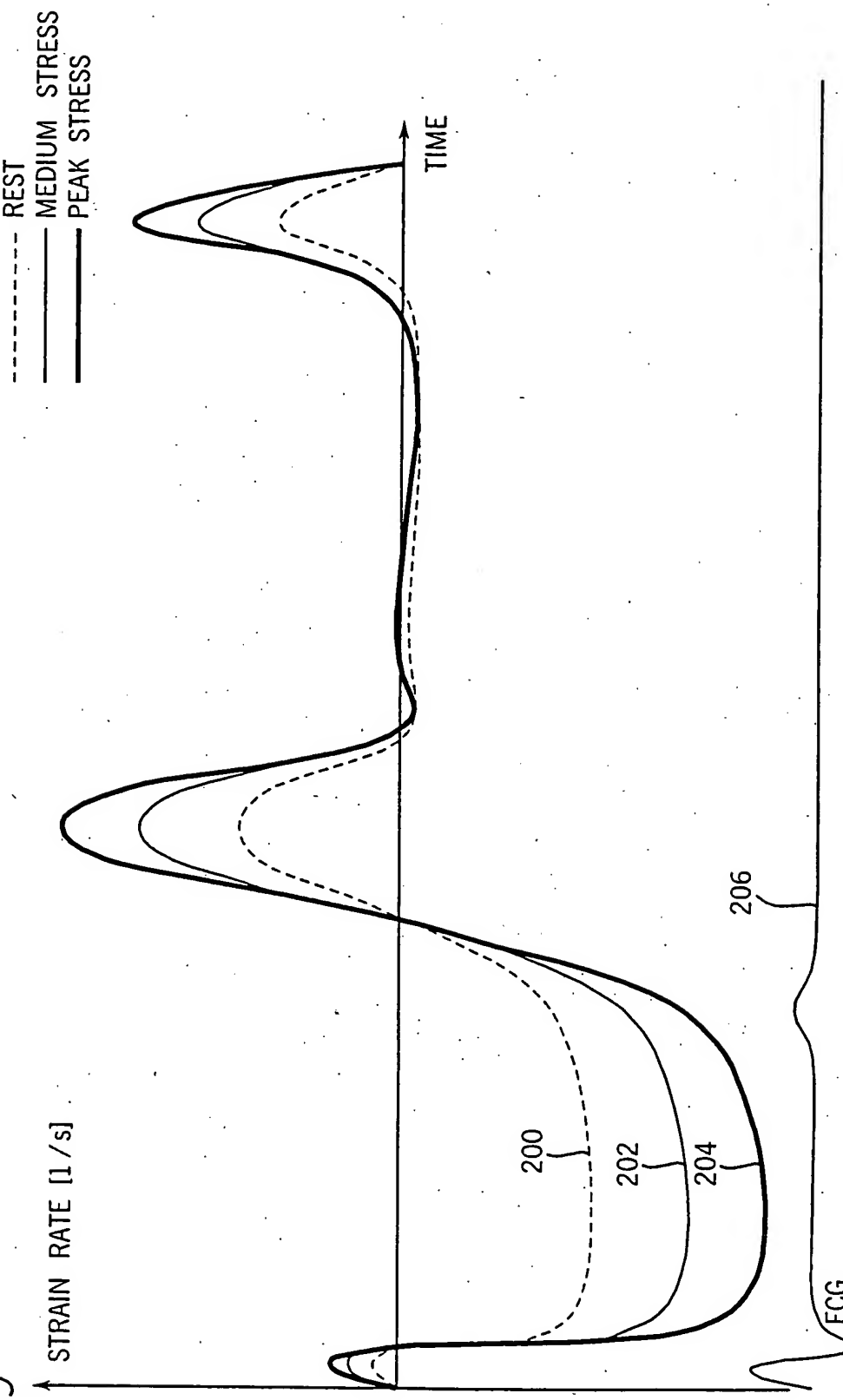

REST
MEDIUM STRESS
PEAK STRESS
TIME
STRAIN RATE [1/s]
200
202
204
206
ECG
e

FIG. 9

FIG. 10

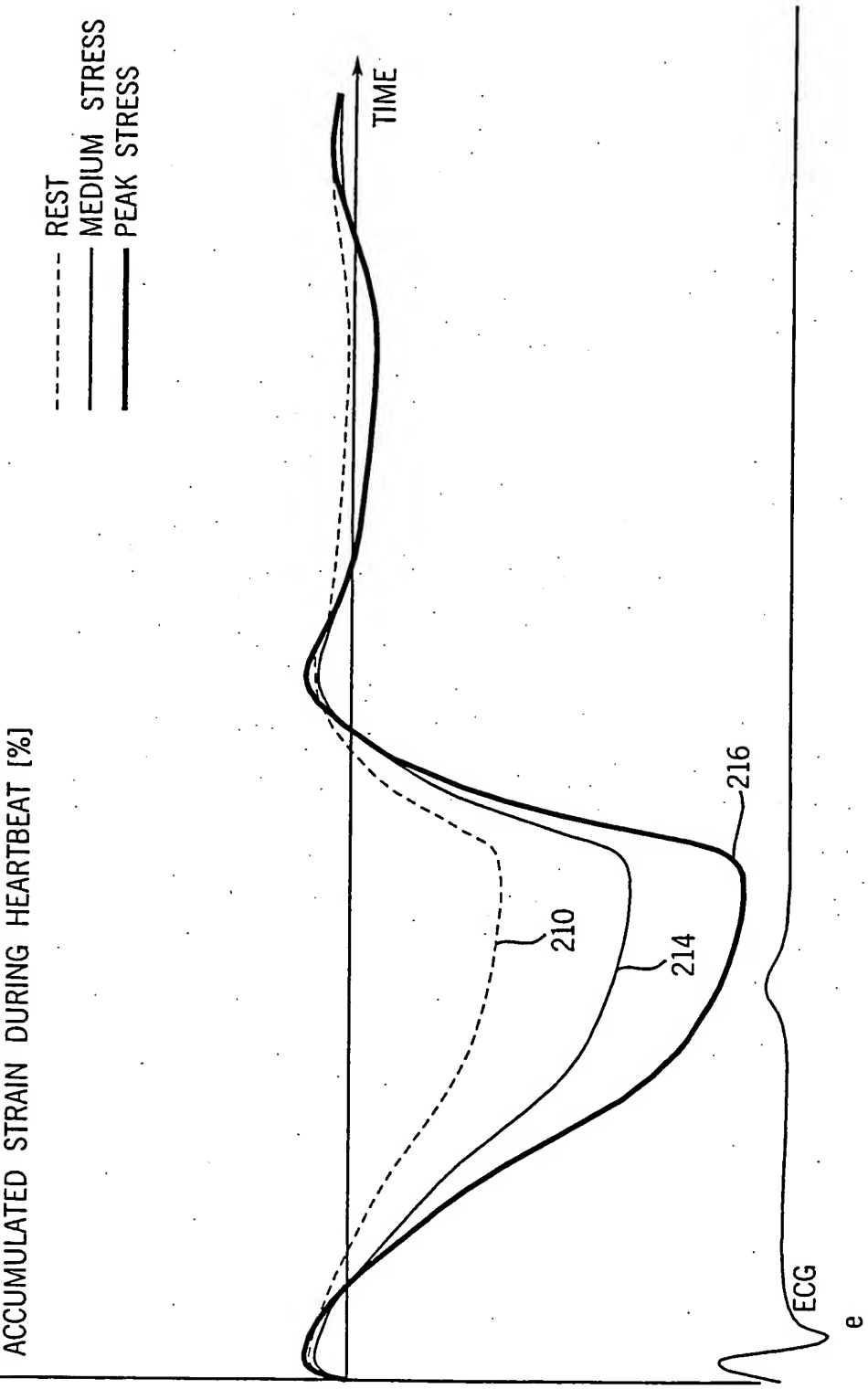

REST
MEDIUM STRESS
PEAK STRESS
ACCUMULATED STRAIN DURING HEARTBEAT [%]
TIME
210
214
216
ECG
e

FIG. 11

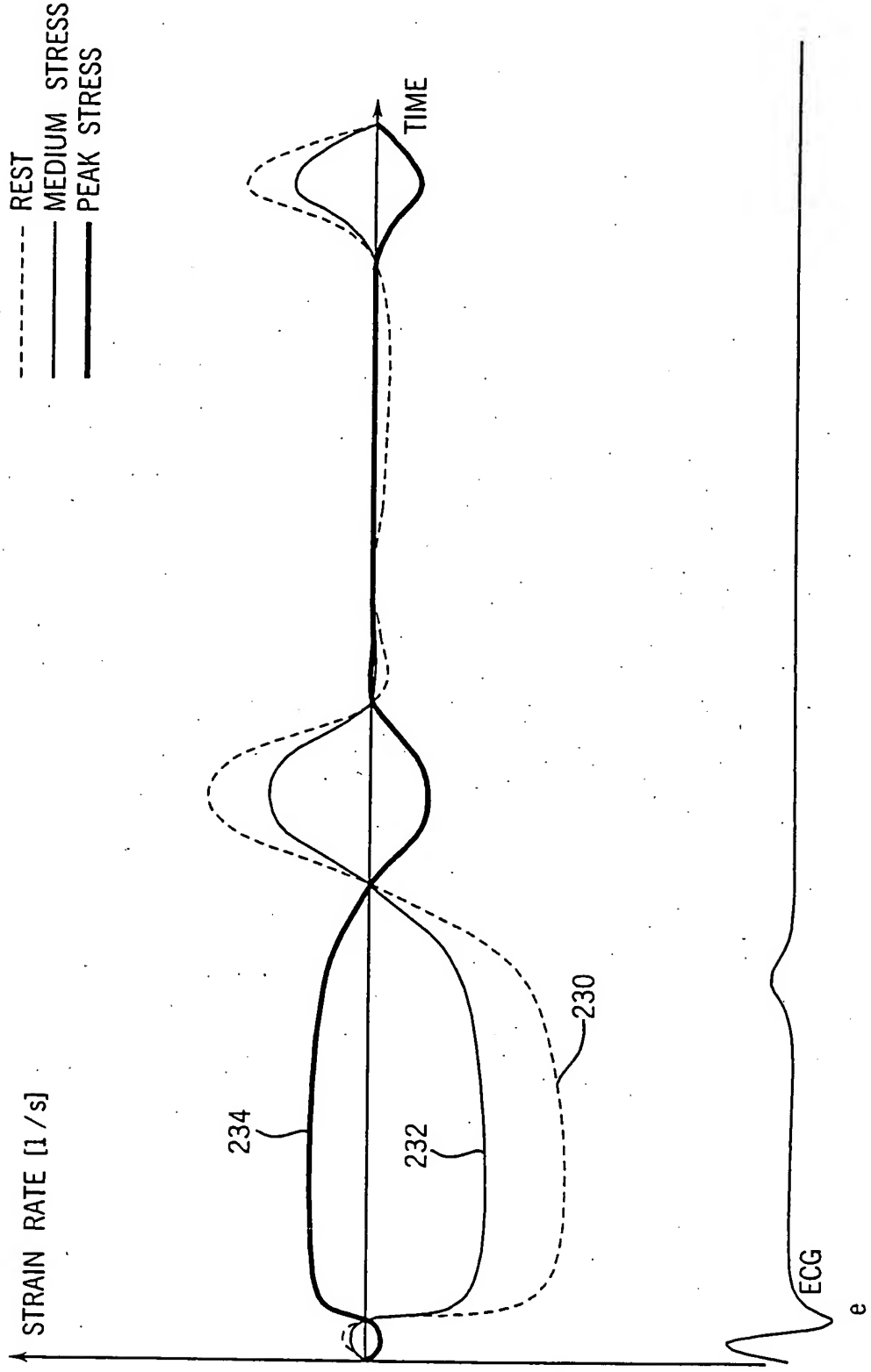
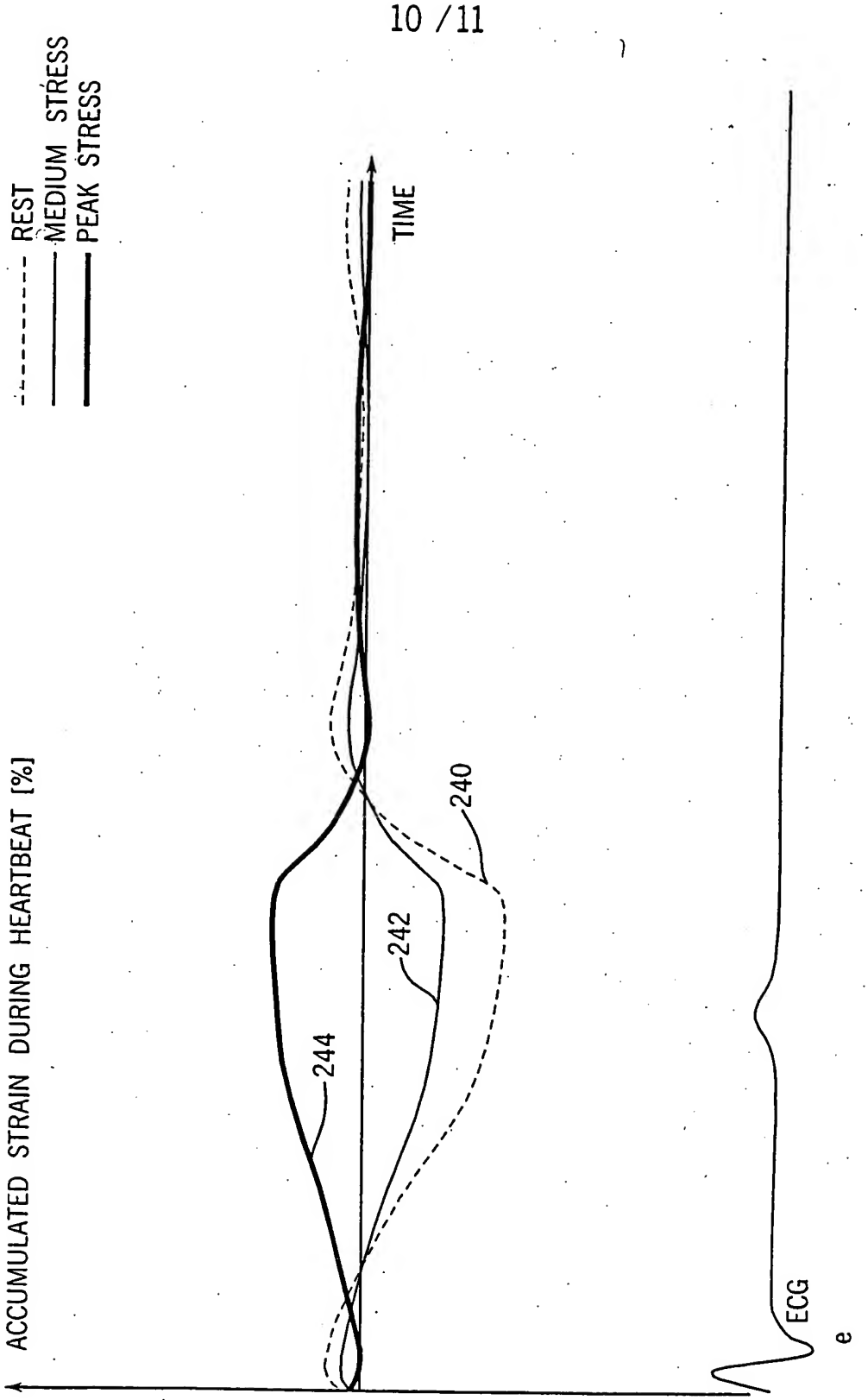

REST
MEDIUM STRESS
PEAK STRESS
TIME
STRAIN RATE [1 / s]
234
232
230
ECG
e

FIG. 12


ACCUMULATED STRAIN DURING HEARTBEAT [%]
REST
MEDIUM STRESS
PEAK STRESS
TIME
240
242
244
ECG
e

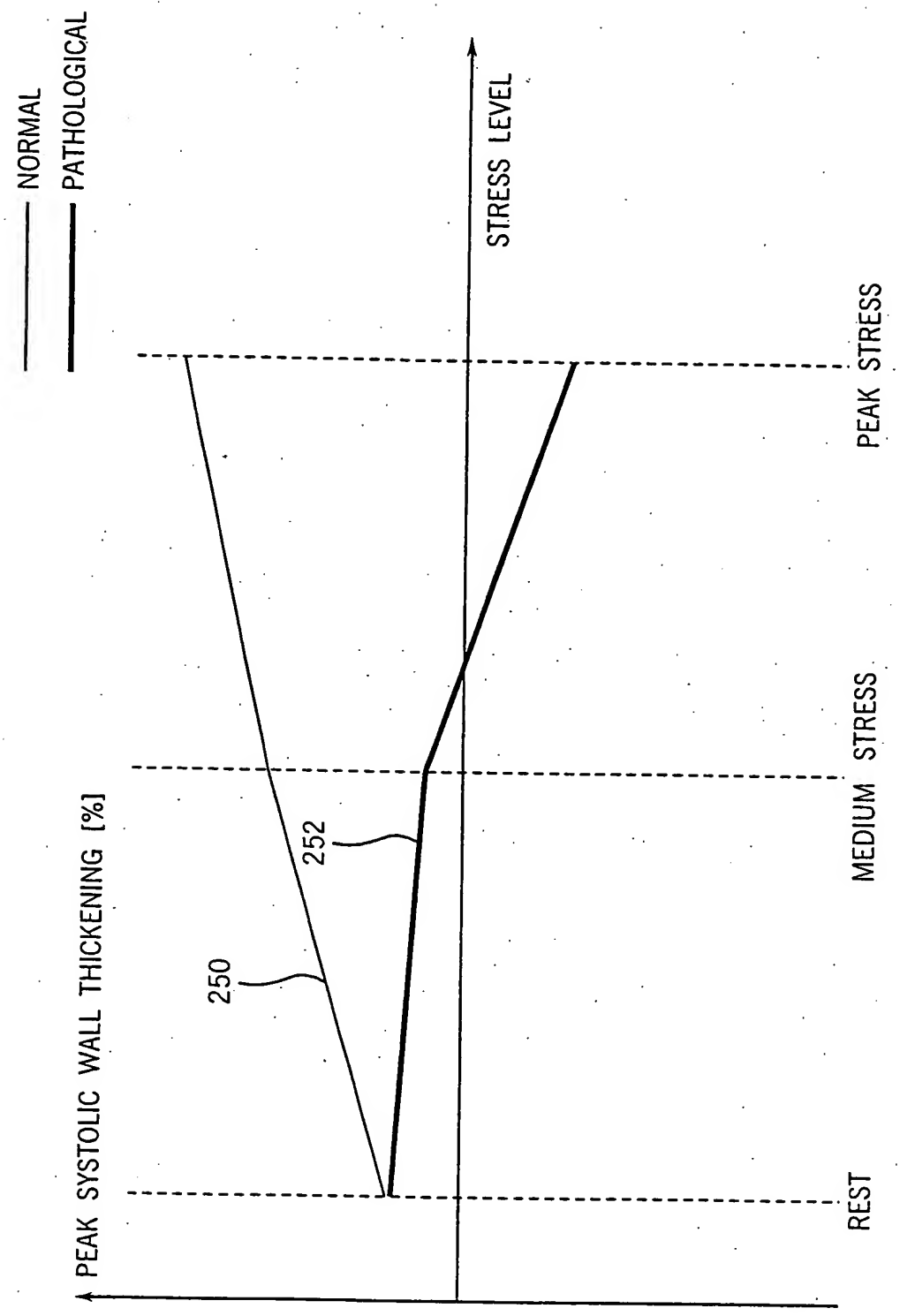

NORMAL
PATHOLOGICAL
STRESS LEVEL
PEAK STRESS
MEDIUM STRESS
REST
PEAK SYSTOLIC WALL THICKENING [%]
250
252

FIG. 13

FEB 19 2004
PATENT & TRADEMARK OFFICE

PTO/SB/05 (06-03)
Approved for use through 07/31/2003. OMB 0651-0032
U.S. Patent and Trademark Office; U.S. DEPARTMENT OF COMMERCE
Under the Paperwork Reduction act of 1995, no persons are required to respond to a collection of information unless it displays a valid OMB control number.

UTILITY PATENT APPLICATION TRANSMITTAL

(Only for new nonprovisional applications under 37 CFR 1.53(b))

Attorney Docket No.	15-DS-000523 DIV2-1 (12483US05)
First Inventor	Torp et al.
Title	METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING
Express Mail Label No.	EV 327683425 US

APPLICATION ELEMENTS

See MPEP chapter 600 concerning utility patent application contents.

ADDRESS TO: **Mail Stop Patent Application, Commissioner for Patents, P.O. Box 1450, Alexandria, VA 22313-1450**

1. ☒ Fee Transmittal Form (e.g., PTO/SB/17) *(Submit an original and a duplicate for fee processing)*
2. ☐ Applicant claims small entity status. See 37 CFR 1.27.
3. ☒ Specification [Total Pages 26] *(preferred arrangement set forth below)*
   - -Descriptive title of the invention
   - -Cross Reference to Related Applications
   - -Statement Regarding Fed sponsored R&D
   - -Reference to sequence listing, a table, or a computer program listing appendix
   - -Background of the Invention
   - -Brief Description of the Drawings *(if filed)*
   - -Detailed Description
   - -Claim(s)
   - -Abstract of the Disclosure
4. ☐ Drawing(s) (35 U.S.C. 113) [Total Sheets _____]
5. Oath or Declaration [Total Sheets _____]
   - a. ☐ Newly executed (original or copy)
   - b. ☒ Copy from a prior application (37 CFR 1.63(d)) *(for continuation/divisional with Box 18 completed)*
     - i ☐ DELETION OF INVENTOR(S) Signed statement attached deleting inventor(s) named in the prior application, see 37 CFR 1.63(d)(2) and 1.33(b).
6. ☐ Application Data Sheet. See 37 CFR 1.76
7. ☐ CD-ROM or CD-R in duplicate, large table or Computer Program (Appendix)
8. Nucleotide and/or Amino Acid Sequence Submission *(if applicable, all necessary)*
   - a. ☐ Computer Readable Form (CRF)
   - b. Specification Sequence Listing on:
     - i. ☐ CD-ROM or CD-R (2 copies); or
     - ii. ☐ Paper
   - c. ☐ Statements verifying identity of above copies

ACCOMPANYING APPLICATION PARTS

9. ☐ Assignment Papers (cover sheet & documents(s))
10. ☐ 37 CFR 3.73(b) Statement *(when there is an assignee)* ☐ Power of Attorney
11. ☐ English Translation Document *(if applicable)*
12. ☒ Information Disclosure Statement (IDS)/PTO-1449 ☐ Copies of IDS Citations
13. ☐ Preliminary Amendment
14. ☒ Return Receipt Postcard (MPEP 503) *(Should be specifically itemized)*
15. ☐ Certified Copy of Priority Document(s) *(if foreign priority is claimed)*
16. ☐ Nonpublication Request under 35 U.S.C. 122(b)(2)(B)(i). Applicant must attach form PTO/SB/35 or its equivalent.
17. ☐ Other: .....

18. If a CONTINUING APPLICATION, *check appropriate box, and supply the requisite information below and in the first sentence of the specification following the title, or in an Application Data Sheet under 37 CFR 1.76:*

☐ Continuation ☒ Divisional ☐ Continuation-in-part (CIP) of prior application No.: This application is a Divisional Application of U.S. Patent Application No. 10/054,033, filed January 22, 2002, now pending; which is a Divisional of U.S. Patent Application No. 09/432,061, filed November 2, 1999, issued as U.S. Patent No. 6,352,507; which is based upon U.S. Provisional Application No. 60/150,265, filed August 23, 1999.

Prior application information: *Examiner:* _____ *Art Unit:* _____

For CONTINUATION OR DIVISIONAL APPS only: The entire disclosure of the prior application, from which an oath or declaration is supplied under Box 5b, is considered a part of the disclosure of the accompanying continuation or divisional application and is hereby incorporated by reference. The incorporation can only be relied upon when a portion has been inadvertently omitted from the submitted application parts.

19. CORRESPONDENCE ADDRESS

☒ Customer Number: 23446 *OR* ☐ Correspondence address below

Name	McAndrews Held & Malloy				
Address	34th Floor				
	500 West Madison Street				
City	Chicago	*State*	IL	*Zip Code*	60661
Country	US	*Telephone*	312/775-8000	*Fax*	312/775-8100

Name (Print/type)	Peter J. McAndrews	*Registration No. (Attorney/Agent)*	38,547
Signature		*Date*	11/10/2003

OIPE
FEB 19 2004
PATENT & TRADEMARK OFFICE
JC102

G.E. DOCKET NUMBER 15DS00523

COMBINED DECLARATION AND POWER OF ATTORNEY FOR PATENT APPLICATION

As a below named inventor, I hereby declare that:

My residence, post office address and citizenship are as stated below next to my name.

I believe I am an original, first and joint inventor of the subject matter which is claimed and for which a patent is sought on the invention entitled

METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING

the specification of which:

X is attached hereto.

was filed on _____ as United States Application Number _____.

I hereby state that I have reviewed and understand the contents of the above identified specification, including the claims, as amended by any amendment referred to above.

I acknowledge the duty to disclose to the United States Patent and Trademark Office all information which I know to be material to patentability as defined in Title 37, Code of Federal Regulations, §1.56.

I hereby claim the benefit under Title 35, United States Code, §119(e) of any United States provisional application(s) listed below.

Application Number	Filing Date
60 / 150,265	August 23, 1999

I hereby appoint the following attorneys and/or agents to prosecute this application and to transact all business in the Patent and Trademark Office connected therewith:

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George P. McAndrews	Reg. No. 22,760
John J. Held	Reg. No. 21,061
Timothy J. Malloy	Reg. No. 25,600
William M. Wesley	Reg. No. 26,521
Lawrence M. Jarvis	Reg. No. 27,341
Gregory J. Vogler	Reg. No. 31,313
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Joseph M. Barich	Reg. No. 42,291
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Address all telephone calls to Dean D. Small at telephone number:

(312) 707-8889.

Address all correspondence to:

Dean D. Small
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34th Floor
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Chicago, Illinois 60661

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This declaration names four inventors below.

Information about first joint inventor:

(given name, family name): Hans Torp

Residence: Arne Lyveien 13
7024 Trondheim

Citizenship: Norway
Post Office Address: Same

First inventor's signature: Hans Torp

Date Signed: 28/10-99

Information about second joint inventor:

(given name, family name): Bjorn Olstad

Residence: _____

_____

Citizenship: Norway
Post Office Address: Same

Second inventor's signature: _____

Date Signed: _____

Information about first joint inventor:

(given name, family name): Hans Torp

Residence: _____

_____

Citizenship: Norway
Post Office Address: Same

First inventor's signature: _____

Date Signed: _____

Information about second joint inventor:

(given name, family name): Bjorn Olstad

Residence: BLEHRSGT. 1

3960 STATHELLE

Citizenship: Norway
Post Office Address: Same

Second inventor's signature: Bjørn Olstad

Date Signed: 28.10.99

Information about third joint inventor:

(given name, family name): Andreas Heimdal

Residence: Klostergt. 74 A
7030 Trondheim, Norway

Citizenship: Norway
Post Office Address: Same

Third inventor's signature: Andreas Heimdal

Date Signed: 29-10-99

Information about fourth joint inventor:

(given name, family name): Steinar Bjaerum

Residence: WEIDEMANNSV. 7A
7014 TRONDHEIM

Citizenship: Norway
Post Office Address: Same

Fourth inventor's signature: Steinar Bjaerum

Date Signed: 28/10-99

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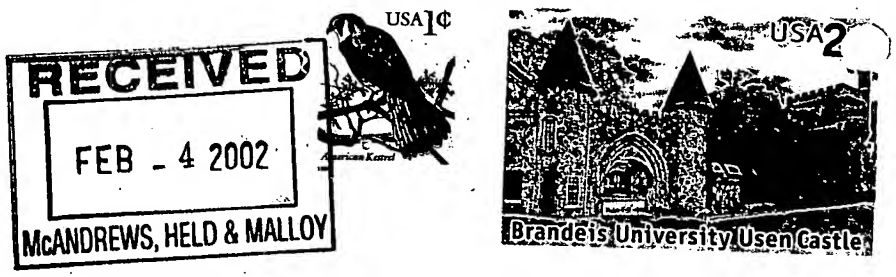

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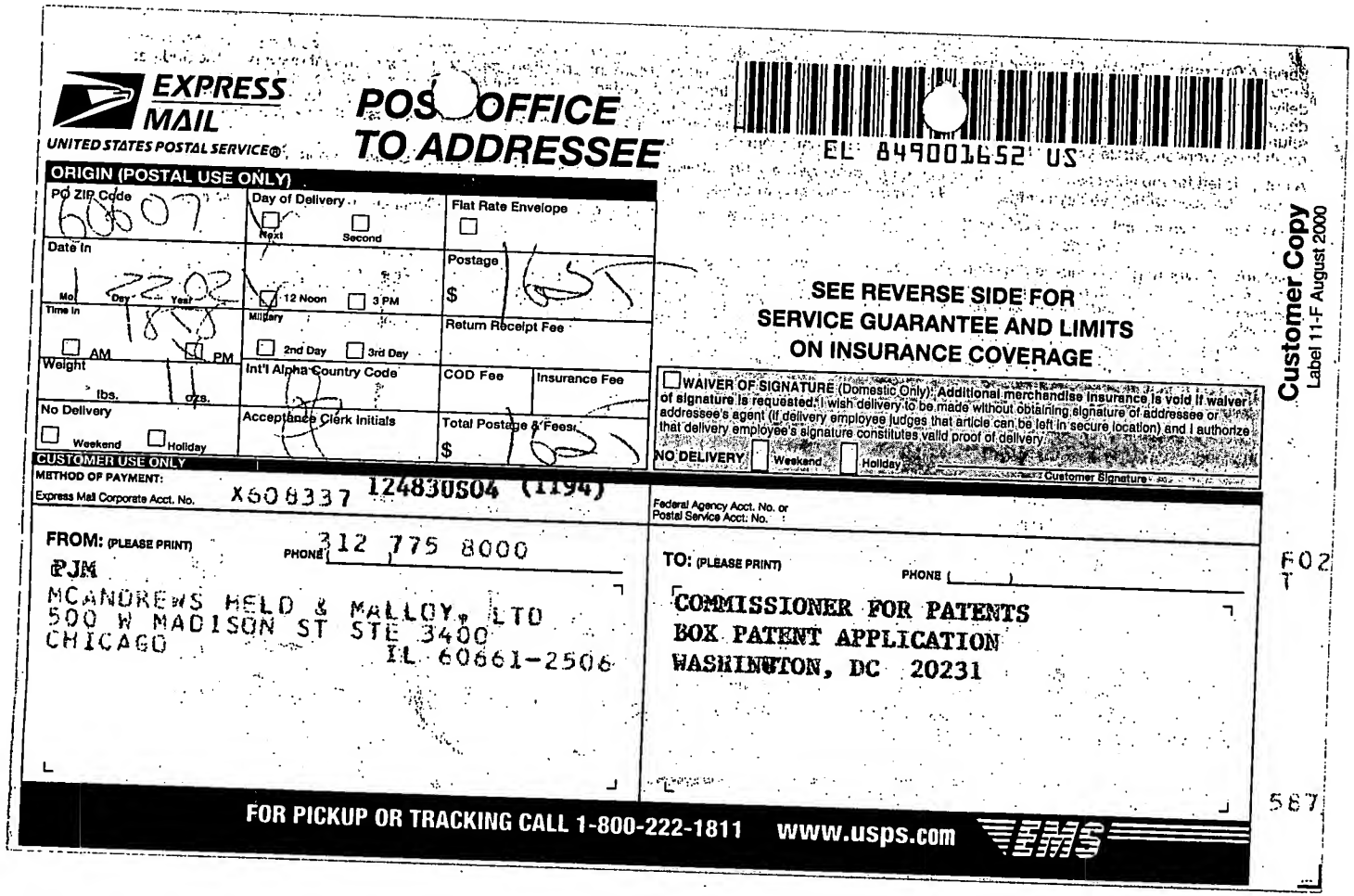

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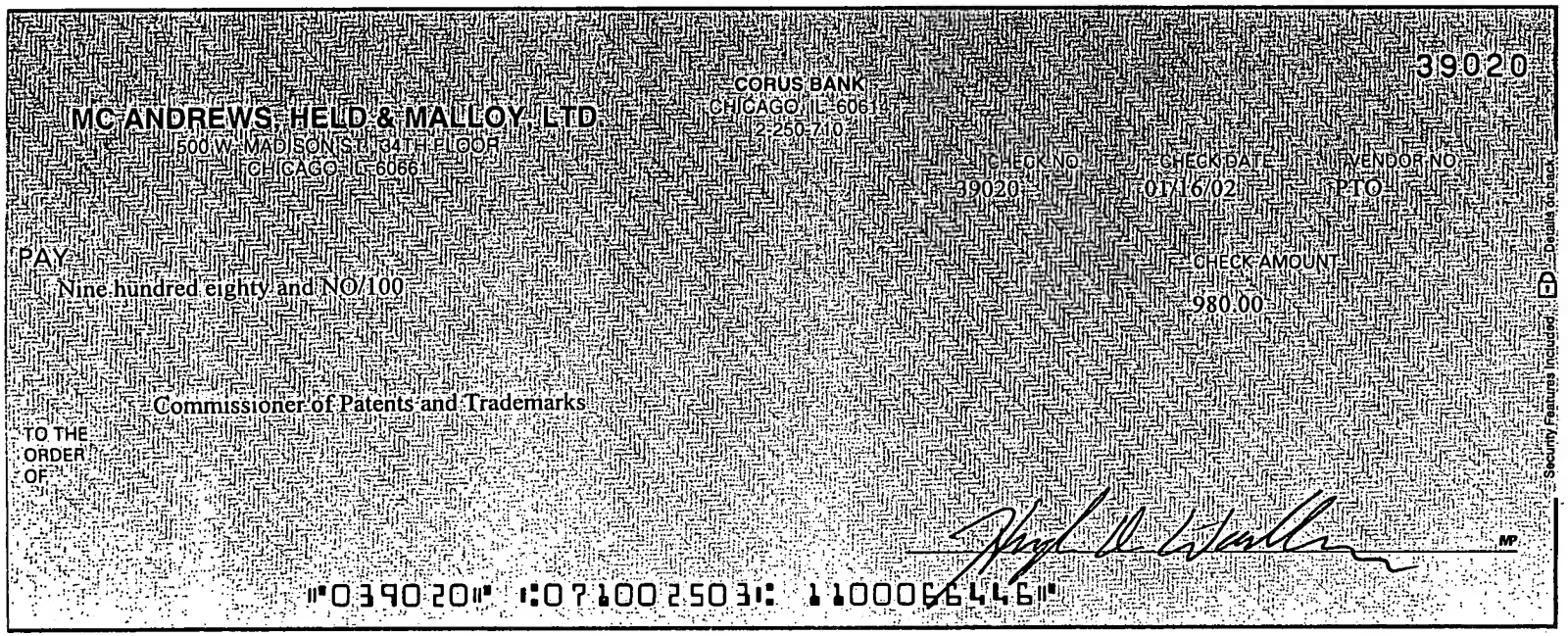

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UTILITY PATENT APPLICATION TRANSMITTAL

(Only for new nonprovisional applications under 37 CFR 1.53(b))

Attorney Docket No.	15-DS-00523 DIV2
First Inventor	Torp et al.
Title	METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING
Express Mail Label No.	EL849001652US

APPLICATION ELEMENTS

See MPEP chapter 600 concerning utility patent application contents.

ADDRESS TO: Commissioner for Patents, Box Patent Application, Washington, DC 20231

1. ☒ Fee Transmittal Form (e.g., PTO/SB/17) *(Submit an original and a duplicate for fee processing)*
2. ☐ Applicant claims small entity status. See 37 CFR 1.27.
3. ☒ Specification [Total Pages 36] *(preferred arrangement set forth below)*
   - -Descriptive title of the invention
   - -Cross Reference to Related Applications
   - -Statement Regarding Fed sponsored R&D
   - -Reference to sequence listing, a table, or a computer program listing appendix
   - -Background of the Invention
   - -Brief Description of the Drawings *(if filed)*
   - -Detailed Description
   - -Claim(s)
   - -Abstract of the Disclosure
4. ☒ Drawing(s) (35 U.S.C. 113) [Total Sheets 11]
5. Oath or Declaration [Total Pages 6]
   - a. ☐ Newly executed (original or copy)
   - b. ☒ Copy from a prior application (37 CFR 1.63(d)) *(for continuation/divisional with Box 18 completed)*
     - i ☐ DELETION OF INVENTORS
       Signed statement attached deleting inventor(s) named in the prior application, see 37 CFR 1.63(d)(2) and 1.33(b).
6. ☐ Application Data Sheet. See 37 CFR 1.76
7. ☐ CD-Rom or CD-R in duplicate, large table or Computer Program (Appendix)
8. Nucleotide and/or Amino Acid sequence Submission *(if applicable, all necessary)*
   - a. ☐ Computer Readable Form (CRF)
   - b. Specification Sequence Listing on:
     - i. ☐ CD-ROM or CD-R (2 copies); or
     - ii. ☐ paper
   - c. ☐ Statements verifying identity of above copies

ACCOMPANYING APPLICATION PARTS

9. ☐ Assignment Papers (cover sheet & documents(s))
10. ☐ 37 CFR 3.73(b) Statement *(when there is an assignee)* ☐ Power of Attorney
11. ☐ English Translation Document *(if applicable)*
12. ☐ Information Disclosure Statement (IDS)/PTO-1449 ☐ Copies of IDS Citations
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14. ☒ Return Receipt Postcard (MPEP 503) *(Should be specifically itemized)*
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16. ☐ Request and Certification under 35 U.S.C. 122(b)(2)(B)(i). Applicant must attach form PTO/SB/35 or its equivalent.
17. ☐ Other:

18. If a CONTINUING APPLICATION, *check appropriate box, and supply the requisite information below and in a preliminary amendment, or in an Application Data Sheet under 37 CFR 1.76:*

☐ Continuation ☒ Divisional ☐ Continuation-in-part (CIP) of prior application No.: 09/432,061

Prior application information: *Examiner: A. Imam* *Group/ARt Unit: 3737*

For CONTINUATION OR DIVISIONAL APPS only: The entire disclosure of the prior application, from which an oath or declaration is supplied under box 5b, is considered a part of the disclosure of the accompanying continuation or divisional application and is hereby incorporated by reference. The incorporation can only be relied upon when a portion has been inadvertently omitted from the submitted application parts.

18. CORRESPONDENCE ADDRESS

☒ Customer Number or Bar Code Label 23446 (Insert Customer No. or Attach bar code label here) or ☒ Correspondence address below

Name	McAndrews Held & Malloy				
Address	34th Floor				
Address	500 West Madison Street				
City	Chicago	State	IL	Zip Code	60661
Country	USA	Telephone	312/775-8000	Fax	312/775-8100

Name (Print/type)	Peter J. McAndrews	Registration No. (Attorney/Agent)	38,547
Signature		Date	01-22-02

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Application Number	Not yet assigned.
Filing Date	Filed herewith.
First Named Inventor	Torp, et al.
Examiner Name	A. Imam
Group Art Unit	3737
Attorney Docket No.	15-DS-00523 DIV2

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101	740	201	370	Utility filing Fee	740.00
106	330	206	165	Design filing Fee	
107	510	207	255	Plant filing fee	
108	740	208	370	Reissue filing fee	
114	160	214	80	Provisional filing fee	

SUBTOTAL (1) ($)740.00

2. EXTRA CLAIM FEES

		Extra Claims	Fee from below	Fee Paid
Total Claims	24 - 20** =	4 x	18.00 =	72.00
Independent Claims	6 - 3** =	2 x	84.00 =	168.00
Multiple Dependent			0 =	0

Large Entity Fee Code	Large Entity Fee ($)	Small Entity Fee Code	Small Entity Fee ($)	Fee Description
103	18	203	9	Claims in excess of 20
102	84	202	42	Independent claims in excess of 3
104	280	204	140	Multiple dependent claim, if not paid
109	84	209	42	**Reissue independent claims over original patent
110	18	210	9	**Reissue claims in excess of 20 and over original patent

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FEE CALCULATION (continued)

3. ADDITIONAL FEES

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127	50	227	25	Surcharge - late provisional filing fee or cover sheet	
139	130	139	130	Non-English specification	
147	2,520	147	2,520	For filing a request for *ex parte* reexamination	
112	920*	112	920*	Requesting publication of SIR prior to Examiner action	
113	1,840*	113	1,840*	Requesting publication of SIR after Examiner action	
115	110	215	55	Extension for reply within first month	
116	400	216	200	Extension for reply within second month	
117	920	217	460	Extension for reply within third month	
118	1,440	218	720	Extension for reply within fourth month	
128	1,960	228	980	Extension for reply within fifth month	
119	320	219	160	Notice of Appeal	
120	320	220	160	Filing a brief in support of an appeal	
121	280	221	140	Request for oral hearing	
138	1,510	138	1510	Petition to institute a public use proceeding	
140	110	240	55	Petition to revive - unavoidable	
141	1,280	241	640	Petition to revive - unintentional	
142	1,280	242	640	Utility issue fee (or reissue)	
143	460	243	230	Design issue fee	
144	620	244	310	Plant issue fee	
122	130	122	130	Petitions to the Commissioner	
123	50	123	50	Processing fee under 37 CFR 1.17(q)	
126	180	126	180	Submission of Information Disclosure Stmt	
581	40	581	40	Recording each patent assignment per property (times number of properties)	
146	740	246	370	Filing a submission after final rejection (37 CFR § 1.129(a))	
149	740	249	370	For each additional invention to be examined (37 CFR 1.129(b))	
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TITLE

METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING

CROSS REFERENCE TO RELATED APPLICATIONS (if applicable)

This application is a Divisional Application of U.S. Patent Application No. 09/432,061, filed November 2, 1999, now pending, which is based upon U.S. Provisional Application No. 60/150,264, filed August 23, 1999.

STATEMENT REGARDING FEDERALLY SPONSORED RESEARCH & DEVELOPMENT (if applicable)

BACKGROUND OF THE INVENTION

The present invention relates to diagnostic ultrasound systems which measure and image anatomical structures, and their movement. More particularly, the present invention relates to a signal processing method and apparatus for calculation and display of tissue deformation to be used in ultrasonic imaging systems.

Recently, within the field of ultrasound imaging, physicians have become interested in using tissue deformation properties, such as tissue strain and strain velocity for clinical measurements.

The term "strain" refers to a characteristic of material being examined. For example, the strain associated with muscle tissue corresponds to a ratio of the change in muscle tissue length during a prescribed time interval to the muscle tissue's initial length. In ultrasound imaging, the rate of change of strain (e.g., strain rate, strain velocity, etc.) may be visually presented to a physician as a colorized 2-dimensional image, where variations in color correspond to different strain velocities.

It has become apparent that the viability of a segment of the muscle is related to the amount of muscle strain and temporal behavior of the strain that is performed by, or is imposed on the muscle segment. Also, it has been determined that malignant tumors may be detected based on their resistance to compression.

One application of real-time strain velocity imaging is in cardiology. The strain velocity gives a direct and quantitative measure for the ability of the myocardium to contract and relax. By imaging along the myocard from an apical view, the local strain velocity component along the long axis of the heart can be measured. Measuring the local strain velocity component gives information about the local shortening and lengthening of the heart wall. By imaging from the parasternal view, the strain velocity component perpendicular to the heart wall can be found. Finding the strain velocity component perpendicular to the heart wall gives information about the local thickening of the muscle. Wall thickening measured with M-mode or from the 2D image is a commonly used measure for muscle viability. With strain velocity imaging, a direct measure for this thickening is available. The strain velocity images can potentially add to the diagnosis of a number of cardiac disorders.

Another application of strain velocity imaging is in heart transplants. Velocity variations inside the myocardium are important for the diagnosis of rejection after heart transplantation. The strain velocity images give a direct display of these velocity variations.

Another application of strain velocity imaging is in noninvasive electrophysiology. The preferred embodiment describes techniques to image the local contraction/relaxation contributions with a high spatial and temporal resolution. Local contraction/relaxation information can be used to accurately determine the localization of, for example, where the mechanical movement in the heart chambers is activated based on a cross section just below the AV-plane. Furthermore, aberrant conduction pathways (Wolf-Parkinson-White) from the atrium to the ventricle can be localized for later ablation. Even the depth inside myocard of these paths can be

better localized with this invention in order to determine if the patient should be treated with catheter techniques or surgical techniques.

Another application of strain velocity imaging is in measuring cardiac wall thickening. A well established methodology in cardiac diagnosis is to acquire a M-Mode image and to measure the wall thickening of myocardium during systole. The preferred embodiment provides techniques to take this wall thickening information and measure it in real-time with a high precision in both the spatial and temporal domain. The high diagnostic relevance of the current wall thickening measurements indicates that the imaging modality described in this invention contains highly relevant information for cardiac diagnosis

To understand strain velocity or strain rate in more detail, it is assumed that an object of initial length L_0 may be stretched or compressed or itself lengthens or contracts to a different length L. The one-dimensional *strain*, defined as

$$\varepsilon = \frac{L - L_0}{L_0} \qquad (1)$$

represents a dimensionless description of the change. If the length L is considered to be a function of time, the temporal derivative of the strain, the strain velocity, can be found using the equation:

$$\dot{\varepsilon} = \frac{\partial \varepsilon}{\partial t} \qquad (2)$$

If the velocity, v of every point in the object is known, an equivalent definition of the strain velocity is:

$$\dot{\varepsilon} = \frac{\partial v}{\partial r} \qquad (3)$$

These equations also provide a useful description of the deformation of the object. In Eq. 3, r is the spatial direction of the stretching or compression. The relation between Eq. 2 and Eq. 3 can be seen if the length L is defined as $L(t) = r_2(t) - r_1(t)$, and $L_0 = L(t_0)$, where r_1 is the distance to one end of the object, and r_2 is the distance to the other, and letting t -> t0, and letting r1 -> r2. As illustrated in Eq. 3, the strain velocity is in fact the spatial gradient of the velocity. The strain velocity thus measures the rate of the deformation of the object. If the strain velocity is zero, the shape of the object is not changing. If the strain velocity is positive, the length of the object is increasing, and if the strain velocity is negative, the length of the object is decreasing. Strain velocity is also known as *rate-of-deformation, stretching, strain rate* or *velocity strain.*

Strain imaging is currently an established research area in ultrasonic imaging. The degree of deformation in imaged structure may be estimated by correlation of 2D images obtained before and after a pressure increase. One disadvantage of estimating image deformation based on correlation of images is that the instantaneous value of the strain is not calculated nor displayed in real-time. The lack of a real-time capability is an important clinical disadvantage. For example, if strain imaging could be performed in real-time, strain imaging could be applied more effectively in cardiac ultrasound or could be used as an interactive inspection modality where anomalies in tissue compressibility can be visualized in real-time according to the pressure gradients that are applied to the imaged structures.

A method of position tracking has been proposed to estimate the local myocardial strain velocity based on radio frequency (RF) M-Mode acquisitions. The position tracking method is described in H. Kanai, H. Hasegawa, N. Chubachi, Y. Koiwa, and M. Tanaka, "Noninvasive evaluation of local myocardial thickening and its color-coded imaging," *IEEE Trans. on Ultrasonics, Ferroelectrics and Frequency Control*, vol. 44, pp. 752-768, 1997. However, the method described in the Kanai et al. article has the disadvantages of poor temporal resolution and high computational cost, which render real-time imaging difficult and costly. Furthermore, the method

described in the Kanai et al. article is a manual M-mode technique, not well suited to form the basis for real-time two-dimensional strain images. Also, the strain velocity is a derivative of a velocity estimate and is therefore very noise sensitive. The fundamental velocity aliasing problem that is inherent in tissue velocity imaging makes noise difficult to overcome because aliasing prevents the pulse repetition frequency from being set at a low enough rate to allow a large observation time. If the observation time could be increased, the noise robustness of the strain velocity images could be significantly improved.

Certain of the above identified difficulties are addressed and overcome according to the teachings of United States Patent Application No. 09/167,896, filed October 7, 1998 and entitled "A METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF STRAIN IN ULTRASOUND IMAGING," which is incorporated herein by reference. However, is an object of the present invention to supplement and/or improve upon such teachings. Certain additional difficulties and shortcomings of the prior art are described below.

To achieve high frame rate in color Doppler applications, two previously known techniques are commonly used: multi line acquisition (MLA) and interleaving. These techniques make it possible to acquire more data than in a basic mode, where the scanner after one pulse is received waits the specified pulse repetition time (T) before firing the next pulse in the same direction. The time to acquire a frame of Doppler data in the basic mode is:

$$t_{D0} = N_b NT, \tag{4}$$

where N is the number of pulses in each direction and N_b is the number of beams in the image. A relatively small extra delay related to the change in setup of the transmitter and beamformer is ignored to simplify the discussion.

In the MLA method, a broad beam is transmitted. When receiving the echo, the signals from all the transducer elements are processed in parallel in two or more beamformers. Each beamformer time delays the element signals differently to

generate different receive beams. This way, two or more beams can be acquired during the time for one pulse-echo cycle, and the frame rate can be increased correspondingly. Using MLA, the time to acquire a frame of Doppler data is

$$t_{DMLA} = \frac{N_b}{N_{MLA}} NT, \tag{5}$$

where N_{MLA} is the number of beams that are processed in parallel.

In the interleaving technique, the waiting time T from one pulse to the next in the same direction is utilized to send pulses in other directions, as illustrated in Fig. 1. There is however a minimum waiting time T_0 where no other pulses can be fired in any direction. This is given by the time for the pulse to travel to the maximum depth and back: $T_0 > 2d/c$. The number of directions that pulses are fired during the time T is called the interleave group size, N_{int}. This obviously has to be an integer number, and $T = N_{int}T_0$. Using interleaving, the time to acquire a frame of Doppler data becomes:

$$t_{D\,\mathrm{int}} = \frac{N_b}{N_{\mathrm{int}} N_{MLA}} NT. \tag{6}$$

Fig. 1 illustrates the pulse order and beam directions in the interleaving method for three different interleave group sizes N_{int}. The number of beams, N_b, equals 8 and packet size, N, equals 2 in the example of Fig. 1. For interleave pattern 100, the interleave group size, N_{int}, equals 8, for interleave pattern 110, N_{int} equals 4 and for interleave pattern 120, N_{int} equals 1.

A typical scanning procedure for a tissue Doppler application is illustrated in Fig. 2. In the example of Fig. 2, the packet size N equals 3 and the interleave group size Nint equals Nb. T is the pulse repetition time, t_T and t_D are the times needed to acquire a tissue frame and a Doppler frame respectively, and t_F is the total acquisition time for one tissue Doppler frame. A tissue frame 130 is first captured, using high beam density. The PRF used for tissue Doppler is usually so low

that only one interleave group is necessary. So N Doppler subframes 132, 134 and 136 are captured separately, usually using fewer beams than in the tissue frame. The velocity is calculated from the N subframes 132, 134 and 136, color coded and then mapped onto the tissue frame. The time to acquire a tissue Doppler frame then becomes:

$$t_F = t_T + \frac{N_b}{N_{MLA}} NT, \tag{7}$$

where t_T is the time required to acquire the tissue frame. It is, thus, apparent that the maximum frame rate is limited by the above described ultrasound data acquisition schemes.

It is known that tissue velocity can be estimated using either the first or the second harmonic component of the ultrasound signal. Using the second harmonic component (*octave* imaging) has been reported to produce an improvement in image quality in gray scale images, and the same improvement can be expected in tissue Doppler. There is however a disadvantage in that the Nyquist limit is halved when using the second harmonic instead of the fundamental component. Using a low PRF is also preferable, since the phase amplitude of the complex signal is increased compared to the noise, resulting in a lower variance in the velocity estimate. A disadvantage of using a low PRF is that the Nyquist limit is further reduced. A reduced Nyquist limit increases the risk of aliasing, which results in the misrepresentation of high velocities.

BRIEF SUMMARY OF THE INVENTION

An ultrasound system and method for calculation and display of tissue deformation parameters are disclosed.

According to one aspect of a preferred embodiment of the present invention, an ultrasound acquisition technique that allows a high frame rate in tissue velocity imaging or strain rate imaging is disclosed. With this acquisition technique the same ultrasound pulses are used for the tissue image and the Doppler based image. A sliding window technique is used for processing.

According to another aspect of a preferred embodiment of the present invention, strain is determined by an accumulation of strain rate estimates for consecutive frames over an interval. The interval may be a triggered interval generated by, for example, an R-wave in an ECG trace. The strain calculation may be improved by moving the sample volume from which the strain rate is accumulated from frame-to-frame according to relative displacement of the tissue within the original sample volume. The relative displacement of the tissue is determined by the instantaneous tissue velocity of the sample volume.

According to another aspect of a preferred embodiment of the present invention, *dr*, the spatial offset used in an estimation of strain rate, is varied adaptively throughout the image. The spatial offset, *dr*, can be maximized to cover the entire tissue segment (e.g., heart wall width) while still keeping both of the sample volumes at each end of the offset within the tissue segment. This may be accomplished by determining whether various parameters (e.g., grayscale value, absolute power estimate, magnitude of the autocorrelation function with unity temporal lag and/or magnitude of strain correlation) of the sample volumes within in the spatial offset are above a given threshold.

According to another aspect of a preferred embodiment of the present invention, a generalized strain rate estimator that is based on a weighted sum of two-sample strain rate estimators with different spatial offsets is employed. The weights are proportional to the magnitude of the strain rate correlation estimate for each spatial offset, and thus reduce the effect of noisy, i.e. poorly correlated, samples.

According to another aspect of a preferred embodiment of the present invention, an improved signal correlation estimator that uses a spatial lag in addition to the usual temporal lag is disclosed. The spatial lag is found from the tissue velocity. The improved signal correlation estimator can be utilized both in the estimation of strain rate and tissue velocity.

According to another aspect of a preferred embodiment of the present invention, tissue velocity is estimated in a manner that reduces aliasing while maintaining spatial resolution. Three copies of a received ultrasound signal are

bandpass filtered at three center frequencies. The middle of the three center frequencies is centered at the second harmonic of the ultrasound signal. A reference tissue velocity is estimated from the two signals filtered at the outside center frequencies. The reference tissue velocity is used to choose a tissue velocity from a number of tissue velocities estimated from the signal centered at the second harmonic.

According to another aspect of a preferred embodiment of the present invention, a method to estimate the strain rate in any direction, not necessarily along the ultrasound beam, based on tissue velocity data from a small region of interest around a sample volume is disclosed.

According to another aspect of a preferred embodiment of the present invention, a plurality of quantitative tissue deformation parameters, such as tissue velocity, tissue velocity integrals, strain rate and/or strain, may be presented as functions of time and/or spatial position for applications such as stress echo. For example, strain rate or strain values for three different stress levels may be plotted together with respect to time over a cardiac cycle. Parameters which are derived from strain rate or strain velocity, such as peak systolic wall thickening percentage, may be plotted with respect to various stress levels.

Other objects, features, and advantages of the present invention will be apparent from the accompanying drawings and from the detailed description that follows below.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 illustrates the pulse order and beam direction for three different interleave group sizes.

Fig. 2 illustrates a typical ultrasound acquisition procedure for a tissue Doppler application.

Fig. 3 illustrates an ultrasound system according to a preferred embodiment of the present invention.

Fig. 4 illustrates an ultrasound acquisition procedure for a tissue Doppler, strain or strain rate application according to a preferred embodiment of the present invention.

Fig. 5 illustrates a graph of the results of a computer simulated comparison of a linear regression strain rate estimator according to the prior art and a weighted strain rate estimator according to a preferred embodiment of the present invention.

Fig. 6 illustrates a graph of the results of a computer simulated comparison of a linear regression strain rate estimator according to the prior art and a weighted strain rate estimator according to a preferred embodiment of the present invention.

Fig. 7 illustrates the coordinates r, u, v, and w, and the insonation angle α used by a strain rate estimate angle correction technique according to a preferred embodiment of the present invention.

Fig. 8 illustrates the velocity components v_v, v_w and v_r, the distance Δr and the angle α in a small muscle segment used by a strain rate estimate angle correction technique according to a preferred embodiment of the present invention.

Fig. 9 illustrates a display of strain rate from multiple stress levels as a function of time for a normal case according to a preferred embodiment of the present invention.

Fig. 10 illustrates a display of accumulated strain from multiple stress levels as a function of time for a normal case according to a preferred embodiment of the present invention.

Fig. 11 illustrates a display of strain rate from multiple stress levels as a function of time for a pathological case according to a preferred embodiment of the present invention.

Fig. 12 illustrates a display of accumulated strain from multiple stress levels as a function of time for a pathological case according to a preferred embodiment of the present invention.

Fig. 13 illustrates a display of a strain derived parameter, peak systolic wall thickening, as a function of stress level according to a preferred embodiment of the present invention.

DETAILED DESCRIPTION OF THE INVENTION

A method and apparatus are described for generating diagnostic images of tissue deformation parameters, such as strain rate, strain and tissue velocity, in real time and/or in a post-processing mode. In the following description, numerous specific details are set forth in order to provide a thorough understanding of the preferred embodiments of the present invention. It will be apparent, however, to one of ordinary skill in the art that the present invention may be practiced without these specific details.

A block diagram for an ultrasound imaging system according to a preferred embodiment of the present invention is shown in Fig. 3. A transmitter 140 drives an ultrasonic transducer 142 to emit a pulsed ultrasonic beam 144 into the body. The ultrasonic pulses are backscattered from structures in the body, like muscular tissue, to produce echoes which return to and are detected by the transducer 142. A receiver 146 detects the echoes. The echoes are passed from the receiver 146 to a complex demodulation stage 148 and a tissue processing stage 149. The complex demodulation stage 148 demodulates the echo signals to form I, Q data pairs representative of echo signals. The demodulated I, Q data pairs are complex Doppler signals that are passed to a tissue deformation calculation stage 150 which carries out tissue velocity, strain rate and/or strain calculations as explained below. The complex Doppler signal is associated with a sample volume defined by a range position and beam in a region of interest. A complex Doppler signal typically comprises a segment of data samples which is used to estimate the Doppler shift. The echo signals are also

passed to the tissue processing stage 149, which performs processing such as B-mode processing to form a 2D or 3D image of the anatomical structure scanned.

The tissue deformation values, e.g., tissue velocity, strain rate and/or strain, output by the tissue deformation calculation stage 150 and the tissue image values output by the tissue processing stage 149 are passed to a display system 152 for display. The display system 152 includes a monitor 154.

United States Patent Application No. 09/167,896, filed October 7, 1998 and entitled "A METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF STRAIN IN ULTRASOUND IMAGING," which is incorporated herein by reference, describes a manner in which a strain rate may be estimated using the system of Fig. 3.

For Strain Rate Imaging (SRI) and other Doppler based applications where a low pulse repetition frequency (PRF) is acceptable, a scanning procedure that allows higher frame rate may be used. Instead of collecting separate tissue frames as illustrated in Fig. 2, the number of beams in the Doppler subframes can be increased to allow tissue visualization based on only these frames. The acquisition of separate tissue frames becomes unnecessary. Fig. 4 illustrates a scanning procedure that allows a high frame rate. This scanning procedure may be used in either tissue Doppler or SRI applications. In the example of Fig. 4, the packet size N = 3 and the interleave group size $N_{int} = N_b$. T is the pulse repetition time, t_T and t_D are the times needed to acquire a tissue frame and a Doppler frame respectively, and t_F is the total acquisition time for one tissue Doppler or SRI frame. As illustrated in Fig. 4, a Doppler frame is still generated from *N* subframes (the subframes are numbered 160, 161, 162, 163 and 164), but a sliding window technique may be used, so the time to produce one Doppler or SRI frame will be only

$$t_{FSRI} = t_T, \tag{8}$$

assuming that the time to acquire one Doppler subframe is equal to the time to acquire one tissue frame in the conventional method. Comparing equations

(7) and (8) one can see that the acquisition time for one frame is greatly reduced and, thus, allowing a higher frame rate.

One parameter that may be calculated by the tissue deformation calculation stage 150 is strain. The relation between the strain and the strain rate can be developed by way of an example. Consider a one-dimensional homogeneous object of length *L(t)* that has a spatially constant strain rate field s(t). The term "strain rate" is here used for the velocity gradient. The velocity field is thus given as:

$$v(t,r) = s(t)r, \tag{9}$$

where r is the position in the object. The velocity at $r = 0$ is set to zero for simplicity, but the same relations will apply also when $v(t,0)$ differs from zero.

The change in length over a small time step Δt can then be estimated as

$$L(t+\Delta t) - L(t) \approx \Delta t s(t) L(t). \tag{10}$$

Letting $\Delta t \to 0$ we get the temporal derivative of the length:

$$\frac{dL(t)}{dt} = \lim_{\Delta t \to 0} \frac{L(t+\Delta t) - L(t)}{\Delta t} = s(t)L(t). \tag{11}$$

The solution to this differential equation is

$$L(t) = L_0 \exp\left(\int_{t_0}^{t} s(\tau) d\tau\right), \tag{12}$$

and the strain is finally found as

$$e(t) = \frac{L(t) - L_0}{L_0} \cdot 100\% = \left[\exp\left(\int_{t_0}^{t} s(\tau) d\tau\right) - 1\right] \cdot 100\%. \tag{13}$$

The strain *e(i)* in a sample volume in the image can be estimated in real-time by replacing the integration in equation ((13) with a cumulative sum:

$$e(i) = [\exp(C(i)) - 1] \cdot 100\%,$$
$$C(i) = C(i-1) + s(i)\Delta t. \qquad (14)$$

Here i is the frame number and Δt is the time between each frame. *C(i)* is the cumulative sum, and *s(i)* is the strain rate estimate for the given sample volume. The accumulation can also be reset at any time, for instance at a specific time trigged by an ECG-signal, by setting *C(i-1)* to zero for the corresponding frame number i. The calculation above can be performed for every sample volume in the image, and the visualization can be performed in the same way as for tissue velocity imageing (TVI) and SRI, only using a color map representing strain rather than tissue velocity or strain rate.

A further improvement is possible if the tissue velocity v is also available for each sample volume. In the cumulative sum for radial sample volume number m_0, the strain rate estimate might then be taken from a different sample volume given by the tissue velocity. First, the frame-to-frame relative displacement index is estimated as

$$d = v\Delta t k_s \qquad (15)$$

where v is the tissue velocity estimate in sample number m_0, and k_s is the spatial sampling frequency. Next, the strain rate estimate from the sample volume number

$$m = m_0 + d \qquad (16)$$

is used in the cumulative sum, rather than m_0. If the tissue movement is only in the direction of the beam, this method allows the cumulative summation to track the motion of the same anatomical sample during its movement. Even if the tissue movement is in other directions, an improvement is expected.

The strain rate estimator in Application No. 09/167,896 was in its simplest form described as:

$$s(r) = (v(r + dr) - v(r)) / dr \qquad (17)$$

where r is the radial position along an ultrasound beam, v is the tissue velocity, and dr is the spatial offset. This spatial offset can be varied adaptively throughout the image. Given an upper and lower limit on the size of dr, it can be increased as much as possible while still keeping both of the sample volumes at each end of the offset within the tissue. There are several different criteria that can be used to ensure that the offset is within the tissue. One possible criteria is that the corresponding tissue sample volumes must have a grayscale value above a given limit. Another possible criteria is that the power estimates of the sample volumes must have absolute values above a given limit. Another possible criteria is that in either of the two sample volumes, the magnitude of the autocorrelation function with unity temporal lag must be above a given limit. Another possible criteria is that the magnitude of the strain correlation (described in equation (8) in Application No. 09/167,896) must be above a given limit. Any of these criteria one can be used separately, or they can be combined so that two or more criteria must be met for a positive determination that the sample volumes at the end of the offset dr are within the tissue.

The tissue deformation calculation stage 14 may calculate strain rate using a strain rate estimator that is based on several samples, and is weighted with the magnitude of a strain correlation estimate. Consider a quadrature demodulated Doppler signal $x(m,n)$, where m is the spatial sample volume index, and n is the temporal index. The signal is assumed to have been acquired using a center frequency f_0, a pulse repetition time T, and a radial sampling frequency r_s equal to the radial size of the point spread function. The speed of sound in the imaged object is assumed to be c. An estimator for strain rate based on M spatial and N temporal samples of the Doppler signal is:

$$s = -\frac{c}{4\pi f_0 Tr_s} \frac{\sum_{m=1}^{M-1} a_m \left|\hat{S}(m)\right| \hat{\omega}_s(m)}{\sum_{m=1}^{M-1} a_m \left|\hat{S}(m)\right|}, \tag{18}$$

where

$$\hat{S}(m) = \frac{1}{M-m} \sum_{k=1}^{M-m} \hat{R}^*(k)\hat{R}(k+m), \tag{19}$$

is the strain rate correlation estimate,

$$\hat{\omega}_s(m) = \frac{1}{\mathrm{m}} \angle \hat{\mathrm{S}}(\mathrm{m}), \tag{20}$$

is the angle of the strain rate correlation estimate, and

$$a_m = m^2\left(1 - \frac{m}{M}\right). \tag{21}$$

is a weighing factor. The signal correlation estimate $\hat{R}(m)$ is described below.

The strain rate estimator of equation (18) has certain advantages over the prior art Myocardial Velocity Gradient (MVG) technique first described in D. Fleming et al., "Myocardial velocity gradients detected by Doppler imaging" Br. J. Radiol., 67(799):679-688, 1994, and further developed by Uematsu et al., "Myocardial velocity gradient as a new indicator of regional left ventricular contraction: Detection by a two-dimensional tissue Doppler imaging technique" J. Am. Col. Cardiol., 26(1):217-23, 1995. For example, Fleming and Uematsu disclose the use of a least squares linear regression of the velocity data to obtain the velocity gradient (strain rate). Linear regression puts equal weight to all the velocity samples. The weighted strain estimator of equation (18), however, uses weights that vary with the magnitude of the strain rate correlation of equation (19) resulting in an improved strain rate estimation.

Figs. 5 and 6 illustrate a computer simulation comparison of a least squares linear regression estimator and the strain rate estimator of equation (18). Fig. 5 illustrates a linear regression fit (dashed line) and a weighted strain rate linear fit (solid line) for simulated velocity estimates (circles) at varying depths. Signals including noise were generated with a velocity gradient (strain rate) of 1.0 s^{-1}. A typical outcome is presented in Fig. 5. Note that the two outermost points give a large error for the linear regression line (dashed line), while the effect on the weighted strain rate estimator is much less. In Fig. 6, strain rates estimated with the linear regression method (stars) and with the weighted strain weight estimator (circles) are compared for 50 independent simulations. Once again, signals including noise were generated with a velocity gradient (strain rate) of 1.0 s^{-1}. The weighted strain rate estimator shows less variance than the linear regression method.

The signal correlation $\hat{R}(m)$ (used in equation (19) above) can be estimated in different ways. For example, one estimate is

$$\hat{R}(m) = \sum_{n=1}^{N-1} x^*(m,n)x(m,n+1). \tag{22}$$

Spatial averaging may also be used to reduce the variance of $\hat{R}(m)$ in equation (22) and other estimators of $\hat{R}(m)$ described herein.

A more robust method to estimate the signal correlation $\hat{R}(m)$ is to introduce a spatial lag Δm, and correlate signal samples from not just the same depth m, but also from $m+\Delta m$:

$$\hat{R}(m) = \sum_{n=1}^{N-1} x^*(m,n)x(m+\Delta m,n+1). \tag{23}$$

The spatial lag Δm preferably is chosen to maximize the magnitude of $\hat{R}(m)$. One way to chose a Δm is through a phase root seeking technique such as described in A. Peasvento and H. Ermert, "Time-efficient and exact algorithms for adaptive temporal stretching and 2D-correlation for elastographic imaging using phase

information" *Proc. of the 1998 Ultrasonic Symposium,* to be published, 1998. Alternatively, the inventors have found that the peak magnitude of $\hat{R}(m)$ is found when the spatial lag Δm is chosen equal to the translation of the tissue from pulse to pulse:

$$\Delta m = \frac{v}{PRF} k_s, \tag{24}$$

where v is the tissue velocity, PRF is the pulse repetition frequency and k_s is the spatial sampling frequency of the signal. This method requires that an un-aliased velocity estimate is available.

The tissue deformation calculation stage 150 may calculate a velocity estimate as follows. Three equal copies of the received signal are band pass filtered with three different filters. Two narrow band filters centered at f_1 and f_2, and a third wider band filter centered at f_3 are used, where $f_1 < f_3 < f_2$, and f_3 is centered around the second harmonic component of the signal. The signal correlation of each of these three signals are estimated using equation (22), resulting in the correlation estimates $\hat{R}_1(m)$, $\hat{R}_2(m)$ and $\hat{R}_3(m)$, respectively. The tissue velocity can be found from the angle of $\hat{R}_3(m)$ as:

$$v_3 = \frac{cPRF}{4\pi f_3} \angle \hat{R}_3(m), \tag{25}$$

where c is the speed of sound. Unfortunately, the velocity estimate of equation (25) is easily aliased. A difference correlation is next found as

$$\hat{R}_d(m) = \hat{R}_1^*(m)\hat{R}_2(m). \tag{26}$$

The velocity of the tissue is found from the angle of this difference correlation as

$$v_d = \frac{cPRF}{4\pi(f_2 - f_1)} \angle \hat{R}_d(m), \qquad (27)$$

where c is the speed of sound. This velocity estimate is not as easily aliased since $(f_2 - f_1) < f_3$. However, with equation (27) the spatial resolution is poor since narrow band signals were used in the estimation of $\hat{R}_1(m)$ and $\hat{R}_2(m)$. To this point, this two-frequency velocity estimation method is similar to a method described in Dousse et al., "Two years experience in measuring velocities beyond the Nyquist limit with Color Flow Mapper" *Proceedings of EURODOP'92*, page 219, Brighton, United Kingdom, 1992.

To regain the spatial resolution of the estimate in equation (25), the following algorithm is used: For each (possibly aliased) velocity estimate v_3, several candidate velocities are found as

$$v_{3,k} = \frac{cPRF}{4\pi f_3}\left(\angle \hat{R}_3(m) + 2k\pi\right), \quad -\left\lfloor \frac{f_3 - (f_2 - f_1)}{2(f_2 - f_1)} \right\rfloor < k < \left\lfloor \frac{f_3 - (f}{2(f_2} \right. \qquad (28)$$

Next, the candidate velocity $v_{3,k}$ that is closest to the (unaliased) difference velocity estimate v_d is chosen as the output velocity estimate. This way, the spatial resolution of the v_3 estimates is kept, while avoiding the problem of aliasing.

A method for angle correction of a strain rate estimation is described with respect to Fig. 7. Locally for each muscle segment of the left cardiac ventricle, the coordinates are defined as:

r - along the ultrasound beam, positive away from the transducer

l – lateral (beam-to-beam), positive from left to right in the ultrasound image

u - circumferential, clockwise seen from the apex

v - meridional (longitudinal), from apex to base

w - transmural, from endo- to epi-card

where u, v and w will be approximately perpendicular, as shown in Fig. 7. The strain rates in these directions are termed s_r, s_l, s_u, s_v and s_w, respectively. The origin (u,v,w)=(0,0,0) does not need to be defined in relation to the macroscopic ventricle geometry, and can be chosen anywhere in the imaged muscle segment.

Furthermore, α is defined as the angle between the v-axis and the r-axis, so that zero degrees corresponds to measuring along the muscle in the meridional direction. It is assumed that the angle α is in the v-w-plane (long axis or apical views), so the problem becomes two-dimensional. Note that the angle α is negative in Fig. 7.

Without loosing generality, it can be assumed that the point (v,w)=(0,0) is not moving. If the strain rate is spatially homogeneous over a small distance Δr in the muscle segment, the muscle point (v,w) will then move with the velocity components:

$$v_v = v s_v \tag{29}$$

and

$$v_w = w s_w . \tag{30}$$

These velocity components are shown in Fig. 8. Fig. 8 is an illustration of the velocity components v_v, v_w and v_r, the distance Δr and the angle α in a small muscle segment. All the parameters are drawn positive, but notice that the angle α is usually negative when imaging from the apex, and that v_v, and consequently v_r, normally are negative during systole. The lateral (beam-to-beam) l-axis is also included for reference. The velocity component along the ultrasound beam in position (v,w) becomes

$$v_r = v s_v \cos\alpha + w s_w \sin\alpha. \tag{31}$$

Notice that the velocity v_r for simplicity is defined positive away from the transducer, i.e., in positive r-direction. This is opposite of the usual definition for the velocity sign in Doppler imaging.

By using velocity-information from more than one beam at a time, it is possible to calculate the strain rate in other directions than along the beam. The beams are assumed to be parallel in the region of interest. The vw-axis system is then a rotation of the lr-axis system by an angle of (α-π/2), and one can write

$$\begin{aligned} v &= r\cos\alpha + l\sin\alpha \\ w &= r\sin\alpha - l\cos\alpha \end{aligned} \tag{32}$$

Substituting these equations in equation (31) one gets

$$v_r = s_v(r\cos\alpha + l\sin\alpha)\cos\alpha + s_w(r\sin\alpha - l\cos\alpha)\sin\alpha \tag{33}$$

Taking the derivatives in the two directions r and l, one gets the two equations

$$\begin{aligned} \frac{\partial v_r}{\partial r} &= s_v\cos^2\alpha + s_w\sin^2\alpha \\ \frac{\partial v_r}{\partial l} &= s_v\sin\alpha\cos\alpha - s_w\sin\alpha\cos\alpha \end{aligned} \tag{34}$$

Solving for s_v and s_w gives

$$\begin{aligned} s_v &= \frac{\partial v_r}{\partial r} + \frac{\partial v_r}{\partial l}\tan\alpha \\ s_w &= \frac{\partial v_r}{\partial r} - \frac{\partial v_r}{\partial l}\cot\alpha \end{aligned} \tag{35}$$

This means the strain rates in the anatomical directions v (meridional) and w (transmural) can be found from the radial and lateral gradients of the measured

radial velocity, as long as the angle α is known. The image plane lr must coincide with the vw plane, which is the case for all apical views and the parasternal long axis view (PLAX). Notice that when imaging from the apex, the angle α will be close to zero for most of the ventricle.

The same formulas apply if one substitutes v with u, so the strain rate in the u-direction (circumferential) can also be found. The image plane lr must then coincide with the uw plane, which is the case for the short axis view (SAX).

There will be some angles where the strain rates are unavailable, though. For the u or the v directions these are the angles where $\tan\alpha$ approaches infinite values. For the w-direction these are the angles where $\cot\alpha$ approaches infinite values.

In the SAX view and using a sector scan, an approximation of α can automatically be found if the user defines the center of the ventricle. By assuming that the SAX cross section of the ventricle is circular, α at a particular location is given as

$$\alpha = \frac{3\pi}{2} - \theta_b + \theta_c \qquad (36)$$

where θ_b is the angle of the ultrasound beam that intersects the point ($\theta_b = 0$ is defined as the center beam), and θ_c is the angle between the center ultrasound beam an imaginary beam from the center of the ventricle through the point.

A preliminary test has been performed using this two-dimensional angle correction method. A velocity data set from a healthy volunteer was obtained using tissue Doppler imaging with high beam density. The short axis view was used and the circumferential and transmural strain rate components in three phases of the cardiac cycle (mid systole, early diastolic relaxation and mid diastole) were estimated. The myocardium was segmented manually. As expected, the resulting images showed that the radial strain rate is equal to the transmural strain rate at twelve and six o'clock, and the circumferential strain rate at two and ten o'clock. Except from where the $\cot\alpha$

or tan α approach infinity, the apparent noise in the images did not seem to be increased by this procedure.

It is also possible to perform three-dimensional angle correction. Locally for each muscle segment of the left cardiac ventricle, the coordinates are defined as:

x – azimuthal (perpendicular to the image plane)

y – lateral (beam-to-beam)

z - along the ultrasound beam

u - circumferential, clockwise seen from the apex

v - meridional (longitudinal), from apex to base

w - transmural, from endo- to epi-card

where the triplets x, y, z and u, v, w locally are assumed to be perpendicular. The strain rates in these directions are termed s_u, s_v and s_w, respectively. The origin (u,v,w)=(0,0,0) does not need to be defined in relation to the macroscopic ventricle geometry, and can be chosen anywhere in the imaged muscle segment.

Without loosing generality, it is assume that the point (u,v,w)=(0,0,0) is not moving. If the strain rate is spatially homogeneous over a small distance Δr in the muscle segment, the muscle point (u,v,w) will then move with the velocity components:

$$v_u = us_u, \quad v_v = vs_v, \quad v_w = ws_w . \tag{37}$$

Using velocity-information from more than one beam at the time it is possible to calculate the strain rate in other directions than along the beam. The beams are assumed to be parallel in the region of interest.

Based on formulas for axis rotation it is possible to express the velocity components in the xyz-directions rather than in the uvw-directions. The velocity component in the z-direction (along the ultrasound beam), v_z, is the one found using Tissue Velocity Imaging. The gradients of this velocity component in each of the three spatial directions are

$$v_{zr} = \frac{\partial v_z}{\partial r}, \quad r = x, y, z \tag{38}$$

The relation to the strain rates in the uvw-directions is

$$\begin{bmatrix} v_{zx} \\ v_{zy} \\ v_{zz} \end{bmatrix} = \mathbf{A}(\alpha, \beta, \gamma) \begin{bmatrix} s_u \\ s_v \\ s_w \end{bmatrix} \tag{39}$$

where $\mathbf{A}(\alpha, \beta, \gamma)$ is a matrix that describes the 3D axis rotation between the uvw-system and the xyz-system, and α, β, and γ are the rotation angles about the u-, v- and w-axis respectively. Except for certain rotation angles, this matrix can be inverted, and the strain rates can be found as:

$$\begin{bmatrix} s_u \\ s_v \\ s_w \end{bmatrix} = \mathbf{A}^{-1}(\alpha, \beta, \gamma) \begin{bmatrix} v_{zx} \\ v_{zy} \\ v_{zz} \end{bmatrix} \tag{40}$$

Estimates for the strain rates in the uvw-directions are found by inserting velocity gradient estimators based on recorded tissue velocity data. Examples of estimators for the velocity gradients are

$$\hat{v}_{zr} = \frac{v_z(r + \Delta r) - v_z(r)}{\Delta r}, \quad r = x, y, z \tag{41}$$

where Δx, Δy, and Δz are the sampling distances in the azimuth, lateral and radial directions in the ultrasound data respectively. Similar methods as described for 1D strain rate can also be used to estimate these velocity gradients, where the radial increment is replaced by an increment in the x- and y-directions.

A further improvement can be achived by performing a least squares fit of the estimated strain rates to the incompressibility equation

$$s_u + s_v + s_w = 0 \tag{42}$$

since muscular tissue can be considered incompressible.

In two dimensions the strain rate estimates reduce to

$$\begin{aligned} s_u &= v_{zz} + v_{zy} \cot \beta \\ s_w &= v_{zz} + v_{zy} \tan \beta \end{aligned} \tag{43}$$

for images in the uw-plane (short axis images) and

$$\begin{aligned} s_v &= v_{zz} + v_{zy} \cot \alpha \\ s_w &= v_{zz} + v_{zy} \tan \alpha \end{aligned} \tag{44}$$

for images in the vw-plane (apical images). There will be some angles where the strain rates are unavailable, though. For the u or the v directions these are the angles where tan approaches infinite values. For the w-direction these are the angles where cot approaches infinite values.

The tissue deformation calculations described herein are suited for quantitative stress echo applications. There are at least four main quantitative parameters that may be extracted, including: tissue velocity, which quantifies wall motion; tissue velocity time integrals which quantify accumulated wall motion during a time interval such as systole; strain rate (velocity gradient), which quantifies the local wall thickening at a given time instant; and strain (integrated strain rate) which quantifies local wall thickening during a time interval such as systole. These parameters are functions of both spatial position and time. From these parameters, other clinically relevant parameters may be derived. One way to present these parameters is to plot pairs of the parameters against each other (similar to pressure-volume-loops). Another useful way to present these parameters is to estimate and record (in a cineloop for example) one or more parameters from different stress levels

in the stress test and then display the respective parameters from the varying stress levels simultaneously.

During a stress echo examination one of the crucial things to assess is segmental wall motion. Typically, the left ventricle is subdivided into segmental areas and a visual assessment of wall motion is done in each of these segments from the various cineloops that are acquired. The 16 segment ASE model of the left ventricle is currently the most common way to subdivide the left ventricle for scoring of stress echo exams. In a visual assessment a given segment is compared in terms of motion and wall thickening by visual comparison of similar views (2-chamber, 4-chamber, lax, sax, etc.) at different stress levels. The stress levels usually include rest, 1 or more levels of stress induced by exercise or pharmacological infusion and finally recovery. A normal reading of a segment is that both wall motion and local wall thickening increase during systole as a function of the applied stress level.

Fig. 9 illustrates how time traces of strain rate for a given location or wall segment can be combined from multiple stress levels. Strain rates estimated during rest (line 200), medium stress (line 202) and peak stress (line 204) are plotted with respect to time. An ECG trace (line 206) is provided for reference at the bottom of the display. A difference in heart rate from the various stress levels is accounted for in this example by time scaling the different strain rate traces. This combined display contains useful clinical information about the local wall function and how the wall segment responds to an increase in stress level. The example is a typical normal reading of longitudinal shortening that can be recorded with an apical view. It should be noted that the longitudinal shortening assessed in this manner also indirectly describes wall thickening in short axis views because of conservation of mass and incompressibility of myocard. The example illustrates a normal reading where longitudinal shortening increases with the stress level.

Fig. 10 is identical to Fig. 9 except that accumulated strain is plotted for rest (line 210), medium stress (line 212) and peak stress (line 214) instead of the instantaneous strain rate. The Fig. 10 demonstrates how the longitudinal shortening increases as a function of stress level. Figs. 11 and 12 correspond to Figs. 9 and 10,

respectively, except that Fig. 11 illustrates a typical pathological reading of strain rates for rest (line 230), medium stress (line 232) and peak stress (line 234), and Fig. 12 illustrates a typical pathological reading of accumulated strain for rest (line 240), medium stress (line 242) and peak stress (line 244). The example of Figs. 11 and 12 illustrates a case with normal rest values for longitudinal shortening, but with a reduction in shortening when the stress level increases. At peak stress the curves illustrate a reverse in both strain rate and strain which can indicate passive stretching of the local wall segment.

Fig. 13 illustrates how characteristic values extracted from the strain information for a given location or wall segment can be displayed as a function of stress level. The example in Fig. 13 is the maximal systolic longitudinal shortening which is plotted as a function of stress level. The normal case (line 250) with a uniform increase in longitudinal shortening and the pathological case (line 252) with a decrease in longitudinal shortening and even a switch to passive stretching during systole are illustrated.

Another useful way to present the quantitative parameters derived from strain is in a Bulls-eye plot by either numerically or graphically labeling each of the areas corresponding to LV segments according to the associated strain derived parameters. The values illustrated in Fig. 13 are examples of such useful strain derived parameters.

In the foregoing specification the invention has been described with reference to specific exemplary embodiments thereof. It will, however, be evident that various modifications and changes may be made thereto without departing from the broader spirit and scope of the invention as set forth in the appended claims. The specification and drawings are, accordingly, to be regarding in an illustrative rather than restrictive sense.

WHAT IS CLAIMED IS:

1. A method for estimating a tissue velocity comprising;

transmitting an ultrasound signal at a fundamental frequency;

receiving an echo of said ultrasound signal from a sample volume and band pass filtering three equal copies of said echo, each copy being filtered in a different band pass range producing a first filtered echo, a second filtered echo and a third filtered echo respectively, said first filtered echo being band pass filtered to a range centered at a first frequency, f_1, that is less than a reference frequency related to said fundamental frequency, said second filtered signal being filtered to a range centered at a second frequency, f_2, that is greater than said reference frequency and said third filtered signal being filtered to a range centered at a third frequency, f_3, that is equal to said reference frequency;

estimating a difference correlation estimate as the product of a complex conjugate of a signal correlation estimate of said first filtered signal and a signal correlation estimate of said second filtered signal;

calculating a first tissue velocity that is proportional to the angle of the difference correlation estimate divided by the difference between said second and first frequencies;

calculating a number of candidate velocities, each of said candidate velocities being proportional to the sum of the angle of a signal correlation estimate of said third filtered signal and a frequency factor divided by the third frequency, where a candidate velocity is calculated for all values of the frequency factor in a range between a negative and positive value of $(f_3 - (f_2 - f_1)) / (2(f_2 - f_1))$; and

selecting one of said candidate velocities that is closest to said first tissue velocity as an output tissue velocity.

2. The method according to claim 1 wherein said reference frequency is equal to a harmonic frequency of said fundamental frequency.

3. A method for estimating a tissue velocity comprising;

transmitting at least two ultrasound pulses along an ultrasound beam;

receiving an echo signal for each of said transmitted ultrasound signals;

bandpass filtering each of said received echo signals with at least two different bandpass filters with center frequencies f_1 and f_2, producing a first filtered echo signal and a second filtered echo signal;

calculating a pulse-to-pulse correlation estimate for each of said filtered echo signals; and

calculating a first tissue velocity that is proportional to the difference in phase angle between the two said correlation estimates, divided by the difference between f_2 and f_1.

4. The method according to claim 3 further comprising:

bandpass filtering said received echo signals with a third bandpass filter with center frequency f_3, producing a third filtered echo signal:

calculating a third pulse-to-pulse correlation estimate for said third filtered echo signal;

calculating a number of candidate velocities, each of said candidate velocities being proportional to the sum of the angle of the correlation estimate for said third filtered echo signal and a frequency factor divided by f_3, where a candidate velocity is calculated for all values of the frequency factor in a range between a negative and positive value of $(f_3 - (f_2 - f_1)) / (2(f_2 - f_1))$; and

selecting one of said candidate velocities that is closest to said first tissue velocity as an output tissue velocity.

5. The method according to claim 4 where said third pulse-to-pulse correlation estimate is calculated with a spatial lag proportional to said first tissue velocity.

6. The method according to claim 4 where at least one of said three bandpass filters are centered in a frequency range generated by nonlinear propagation of said transmitted pulses.

7. A method for performing quantitative stress echo ultrasound comprising:

estimating and storing a tissue deformation value for a heart wall tissue segment of a patient over a cardiac interval during each of at least two stress periods, where a level of stress on the patient is different for each of said at least two stress periods; and

simultaneously displaying the estimated strain rates for each of said at least two stress periods as a function of time over the cardiac interval.

8. The method according to claim 7 wherein the cardiac interval corresponds to an R to R interval of the cardiac cycle.

9. The method according to claim 7 wherein the display of strain rates for at least one of said at least two stress periods is time scaled such that the

length of the cardiac interval during each of said at least two stress periods appears to be equal in length.

10. The method according to claim 7 wherein the tissue deformation value is strain rate.

11. The method according to claim 7 wherein the tissue deformation value is strain accumulated over said cardiac interval.

12. The method according to claim 7 wherein the at least two stress periods comprises three stress periods.

13. A method for generating tissue deformation information comprising:

acquiring echo signals for a plurality of beams and a plurality of range positions along ultrasonic beams in an area of interest to cover a spatial region;

determining a beam angle between the ultrasonic beams and a principle direction for local tissue deformation;

computing at least one angle corrected tissue deformation parameter along said principal direction for at least one spatial location; and

displaying at least one of the said angle corrected tissue deformation parameters on a display unit.

14. The method according to claim 13 wherein said ultrasonic beams are generated with a high lateral resolution inside said area of interest.

15. The method according to claim 13 wherein the said beam angle determination is computed based on a direction along and perpendicular to a user defined polygon.

16. The method according to claim 13 wherein the said computation of at least one angle corrected tissue deformation parameter comprises:

computing a radial velocity gradient radially along the ultrasound beam;

computing a lateral velocity gradient laterally between beams at a fixed range location; and

deriving angle corrected tissue deformation parameters as a linear combination of said radial and lateral velocity gradients determined by said beam angle.

17. The method according to claim 16 further comprising:

spatially averaging said radial and lateral velocity gradients.

18. The method according to claim 13 wherein changes in at least one of said angle corrected tissue deformation parameters is displayed as a function of time for a given anatomical location.

19. The method according to claim 15 wherein the said display is a M-Mode display displaying at least one of said angle corrected tissue deformation parameters with time versus location on said user defined polygon.

20. A method for real-time imaging of temporally accumulated a tissue movement property comprising:

acquiring echo signals for a plurality of ultrasonic beams and a plurality of range positions along said ultrasonic beams in an area of interest to cover a spatial region;

estimating at least one tissue movement property inside the said area of interest;

obtaining a sequence of trigger events;

accumulating said tissue movement property estimates since a most recent trigger event for said spatial region into an accumulation image; and

displaying said accumulation image on a display unit in real-time.

21. The method according to claim 20 wherein said tissue movement property is tissue velocity and said accumulation image is distance computed as a time velocity integral since the most recent trigger event.

22. The method according to claim 20 wherein the said tissue movement property is strain rate and the associated said accumulation image is strain computed as a monotonic map of the sum of strain rate estimates since the most recent trigger event.

23. The method according to claim 20 wherein the said trigger events identify an R event in the cardiac cycle.

24. The method according to claim 20 further comprising:

measuring tissue velocity along said ultrasound beams for all range positions inside said area of interest; and

accumulation tissue velocity from different range positions for each temporally successive frame in order to compensate for movement of anatomical locations along the ultrasound beam measured by said tissue velocity measurement.

METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING

ABSTRACT OF THE DISCLOSURE

An ultrasound system and method for calculation and display of tissue deformation parameters are disclosed. An ultrasound acquisition technique that allows a high frame rate in tissue velocity imaging or strain rate imaging is employed. With this acquisition technique the same ultrasound pulses are used for the tissue image and the Doppler based image. A sliding window technique is used for processing. The tissue deformation parameter strain is also determined by an accumulation of strain rate estimates for consecutive frames over an interval. The interval may be a triggered interval generated by, for example, an R-wave in an ECG trace. The strain calculation may be improved by moving the sample volume from which the strain rate is accumulated from frame-to-frame according to the relative displacement of the tissue within the original sample volume. The relative displacement of the tissue is determined by the instantaneous tissue velocity of the sample volume. An estimation of strain rate based upon a spatial derivative of tissue velocity is improved by adaptively varying the spatial offset, *dr*. The spatial offset, *dr*, can be maximized to cover the entire tissue segment (e.g., heart wall width) while still keeping both of the sample volumes at each end of the offset within the tissue segment. This may be accomplished by determining whether various parameters (e.g., grayscale value, absolute power estimate, magnitude of the autocorrelation function with unity temporal lag and/or magnitude of strain correlation) of the sample volumes within in the spatial offset are above a given threshold. Strain rate may be estimated using a generalized strain rate estimator that is based on a weighted sum of two-sample strain rate estimators with different spatial offsets. The weights are proportional to the magnitude of the strain rate correlation estimate for each spatial offset, and thus reduce the effect of noisy, i.e. poorly correlated, samples. An improved signal correlation estimator that uses a spatial lag in addition to the usual

temporal lag is disclosed. The spatial lag is found from the tissue velocity. The improved signal correlation estimator can be utilized both in the estimation of strain rate and tissue velocity. Tissue velocity may be estimated in a manner that reduces aliasing while maintaining spatial resolution. Three copies of a received ultrasound signal are bandpass filtered at three center frequencies. The middle of the three center frequencies is centered at the second harmonic of the ultrasound signal. A reference tissue velocity is estimated from the two signals filtered at the outside center frequencies. The reference tissue velocity is used to choose a tissue velocity from a number of tissue velocities estimated from the signal centered at the second harmonic. A method to estimate the strain rate in any direction, not necessarily along the ultrasound beam, based on tissue velocity data from a small region of interest around a sample volume is disclosed. Quantitative tissue deformation parameters, such as tissue velocity, tissue velocity integrals, strain rate and/or strain, may be presented as functions of time and/or spatial position for applications such as stress echo. For example, strain rate or strain values for three different stress levels may be plotted together with respect to time over a cardiac cycle. Parameters which are derived from strain rate or strain velocity, such as peak systolic wall thickening percentage, may be plotted with respect to various stress levels.

G.E. DOCKET NUMBER 15DS00523

COMBINED DECLARATION AND POWER OF ATTORNEY FOR PATENT APPLICATION

As a below named inventor, I hereby declare that:

My residence, post office address and citizenship are as stated below next to my name.

I believe I am an original, first and joint inventor of the subject matter which is claimed and for which a patent is sought on the invention entitled

METHOD AND APPARATUS FOR PROVIDING REAL-TIME CALCULATION AND DISPLAY OF TISSUE DEFORMATION IN ULTRASOUND IMAGING

the specification of which:

X is attached hereto.

was filed on _____ as United States Application Number _____.

I hereby state that I have reviewed and understand the contents of the above identified specification, including the claims, as amended by any amendment referred to above.

I acknowledge the duty to disclose to the United States Patent and Trademark Office all information which I know to be material to patentability as defined in Title 37, Code of Federal Regulations, §1.56.

I hereby claim the benefit under Title 35, United States Code, §119(e) of any United States provisional application(s) listed below.

Application Number	Filing Date
60 / 150,265	August 23, 1999

I hereby appoint the following attorneys and/or agents to prosecute this application and to transact all business in the Patent and Trademark Office connected therewith:

Christian G. Cabou	Reg. No. 35,467
Phyllis Y. Price	Reg. No. 34,234
Ronald E. Myrick	Reg. No. 26,315
Henry J. Policinski	Reg. No. 26,621
Jay L. Chaskin	Reg. No. 24,030
George P. McAndrews	Reg. No. 22,760
John J. Held	Reg. No. 21,061
Timothy J. Malloy	Reg. No. 25,600
William M. Wesley	Reg. No. 26,521
Lawrence M. Jarvis	Reg. No. 27,341
Gregory J. Vogler	Reg. No. 31,313
Jean Dudek Kuelper	Reg. No. 30,171
Herbert D. Hart III	Reg. No. 30,063
Robert W. Fieseler	Reg. No. 31,826
D. David Hill	Reg. No. 35,543
Thomas J. Wimbiscus	Reg. No. 36,059
Steven J. Hampton	Reg. No. 33,707
Priscilla F. Gallagher	Reg. No. 32,223
Stephen F. Sherry	Reg. No. 30,590
Patrick J. Arnold Jr.	Reg. No. 37,769
George Wheeler	Reg. No. 28,766
Janet M. McNicholas	Reg. No. 32,918
Ronald E. Larson	Reg. No. 24,478
Christopher C. Winslade	Reg. No. 36,308
Edward A. Mas II	Reg. No. 37,179
Gregory C. Schodde	Reg. No. 36,668
Edward W. Remus	Reg. No. 25,703
Donald J. Pochopien	Reg. No. 32,167
Sharon A. Hwang	Reg. No. 39,717
David D. Headrick	Reg. No. 40,642
Dean D. Small	Reg. No. 34,730
Alejandro Menchaca	Reg. No. 34,389
Kirk A. Vander Leest	Reg. No. 34,036
Richard T. McCaulley, Jr.	Reg. No. 41,977
Anthony E. Dowell	Reg. No. 39,661
Peter J. McAndrews	Reg. No. 38,547
Krista S. Schwartz	Reg. No. 42,134
Eligio C. Pimentel	Reg. No. 42,076
John F. Nethery	Reg. No. 42,928
Paul L. Rodriguez	Reg. No. 40,788
Stephen H. Bean	Reg. No. 41,607
James P. Murphy	Reg. No. 40,741
Joseph M. Barich	Reg. No. 42,291
Scott P. McBride	Reg. No. 42,853

Address all telephone calls to Dean D. Small at telephone number:

(312) 707-8889.

Address all correspondence to:

Dean D. Small
McAndrews, Held & Malloy, Ltd.
34th Floor
500 W. Madison Street
Chicago, Illinois 60661

I hereby declare that all statements made herein of my own knowledge are true and that all statements made on information and belief are believed to be true; and further that these statements were made with the knowledge that willful false statements and the like so made are punishable by fine or imprisonment, or both, under Section 1001 of Title 18 of the United States Code and that such willful false statements may jeopardize the validity of the application or any patent issued thereon.

This declaration names four inventors below.

Information about first joint inventor:

(given name, family name): Hans Torp

Residence: Ameliyveien 13
7024 Trondheim

Citizenship: Norway
Post Office Address: Same

First inventor's signature: Hans Torp

Date Signed: 28/10-99

Information about second joint inventor:

(given name, family name): Bjorn Olstad

Residence: _____

_____

Citizenship: Norway
Post Office Address: Same

Second inventor's signature: _____

Date Signed: _____

Information about first joint inventor:

(given name, family name): Hans Torp

Residence: _____

_____

Citizenship: Norway
Post Office Address: Same

First inventor's signature: _____

Date Signed: _____

Information about second joint inventor:

(given name, family name): Bjorn Olstad

Residence: BLEHRSGT. 1

3960 STATHELLE

Citizenship: Norway
Post Office Address: Same

Second inventor's signature: Bjørn Olstad

Date Signed: 28.10.99

Information about third joint inventor:

(given name, family name): Andreas Heimdal

Residence: Klostergt. 74 A
7030 Trondheim, Norway

Citizenship: Norway
Post Office Address: Same

Third inventor's signature: Andreas Heimdal

Date Signed: 29-10-99

Information about fourth joint inventor:

(given name, family name): Steinar Bjaerum

Residence: WEIDEMANNSV. 7A
7014 TRONDHEIM

Citizenship: Norway
Post Office Address: Same

Fourth inventor's signature: Steinar Bjarum

Date Signed: 28/10-99

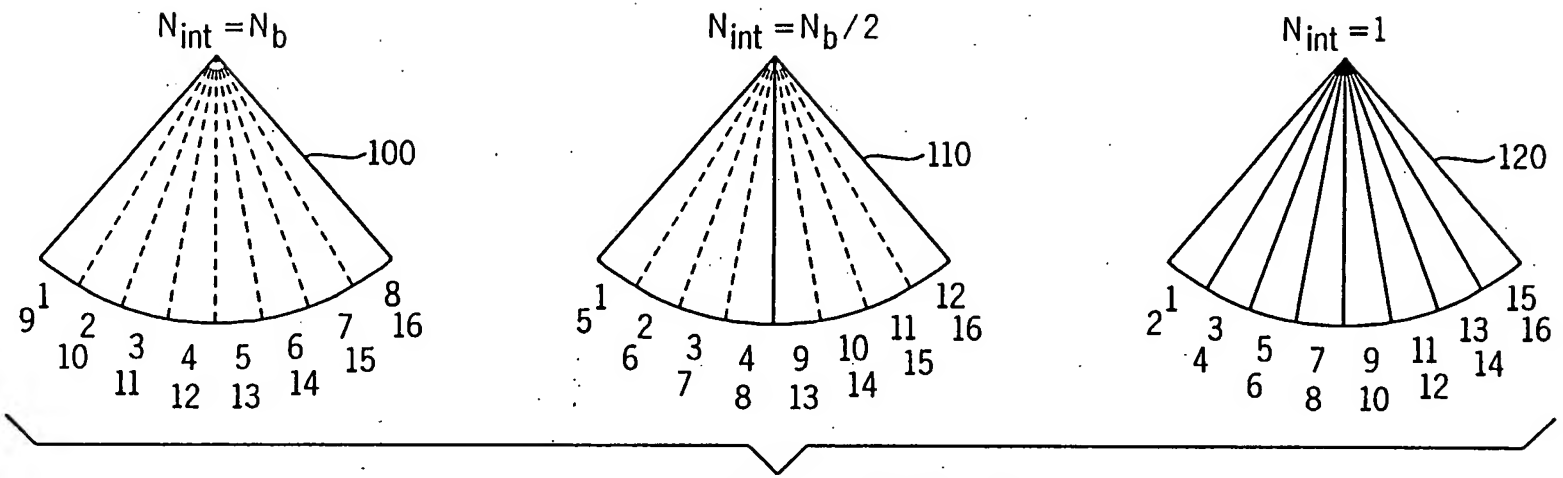


FIG. 1 (PRIOR ART)

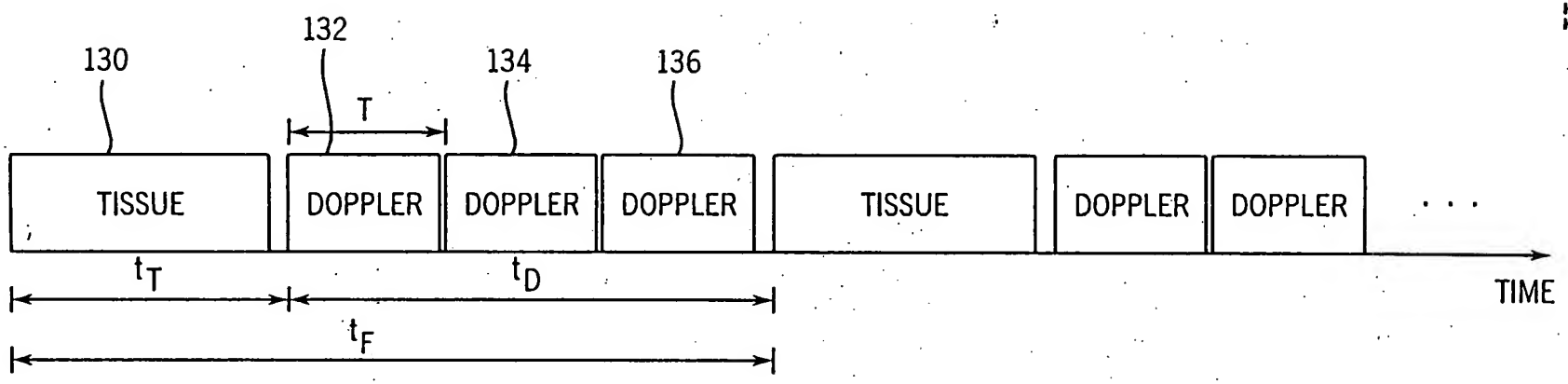


FIG. 2 (PRIOR ART)

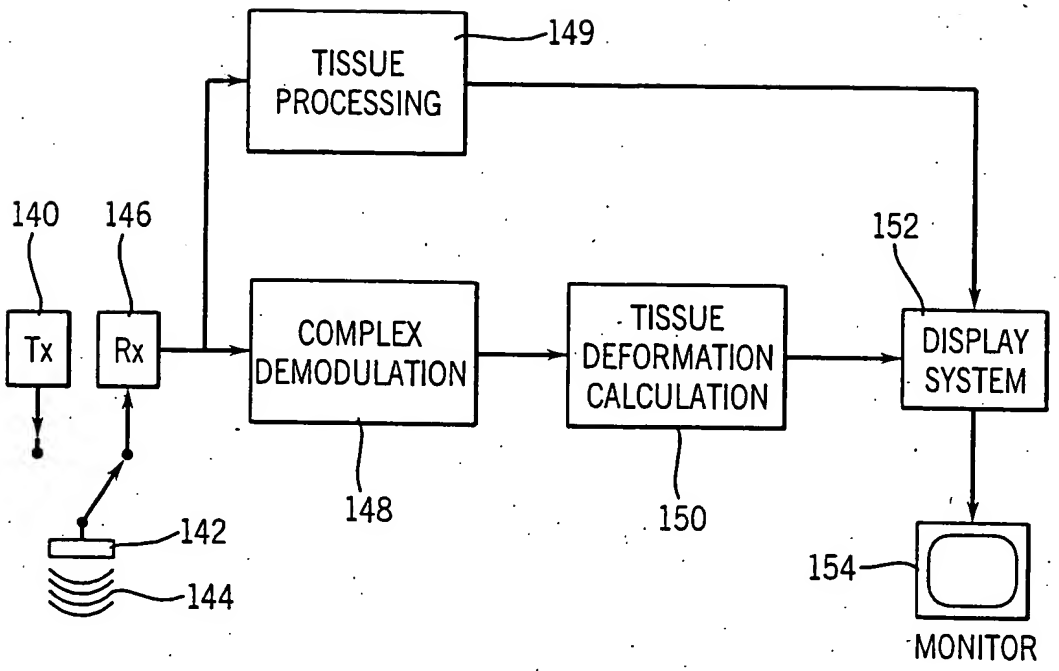

TISSUE PROCESSING
149
140
146
152
Tx
Rx
COMPLEX DEMODULATION
TISSUE DEFORMATION CALCULATION
DISPLAY SYSTEM
148
150
142
144
154
MONITOR

FIG. 3

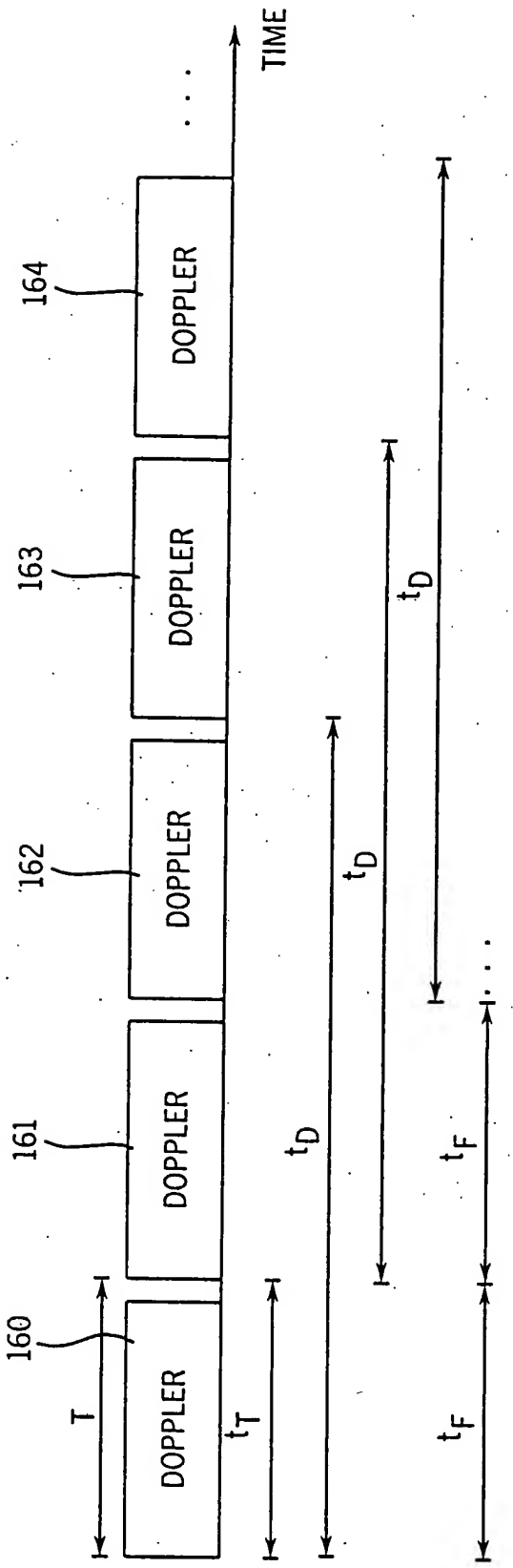


FIG. 4

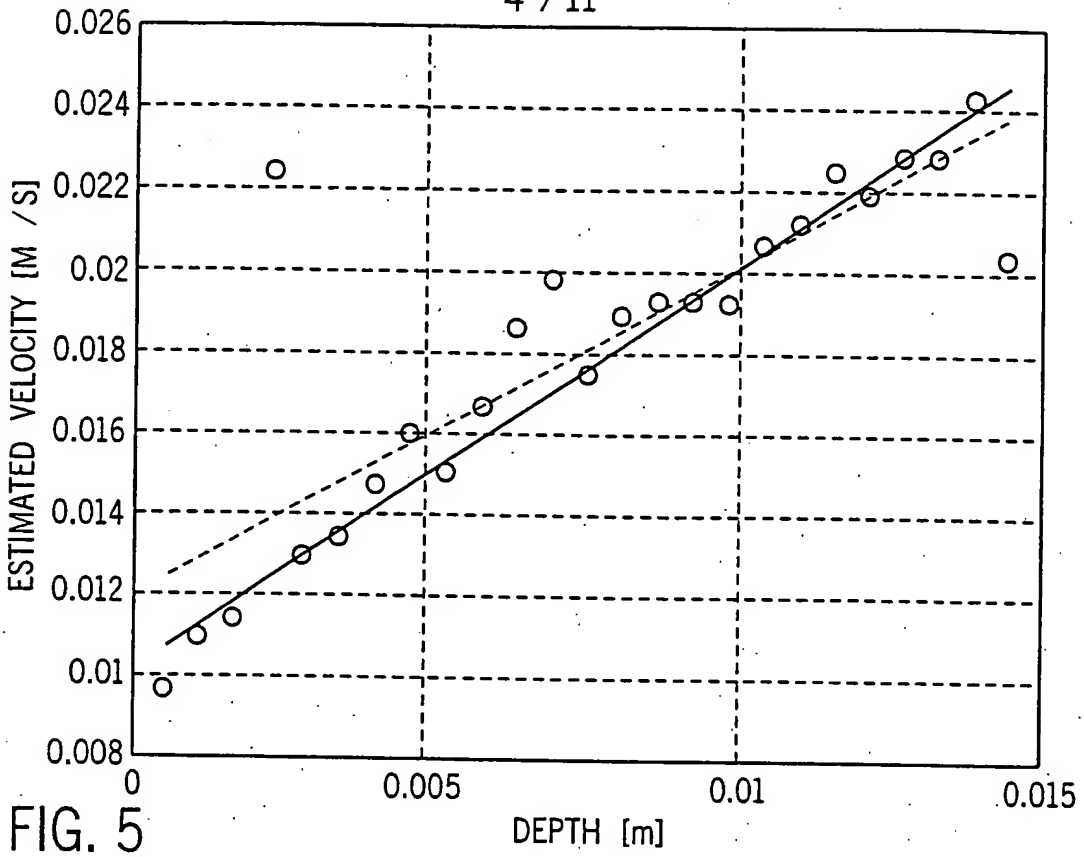

0.026
0.024
0.022
0.02
0.018
0.016
0.014
0.012
0.01
0.008
ESTIMATED VELOCITY [M /S]
0
0.005
0.01
0.015
DEPTH [m]

FIG. 5

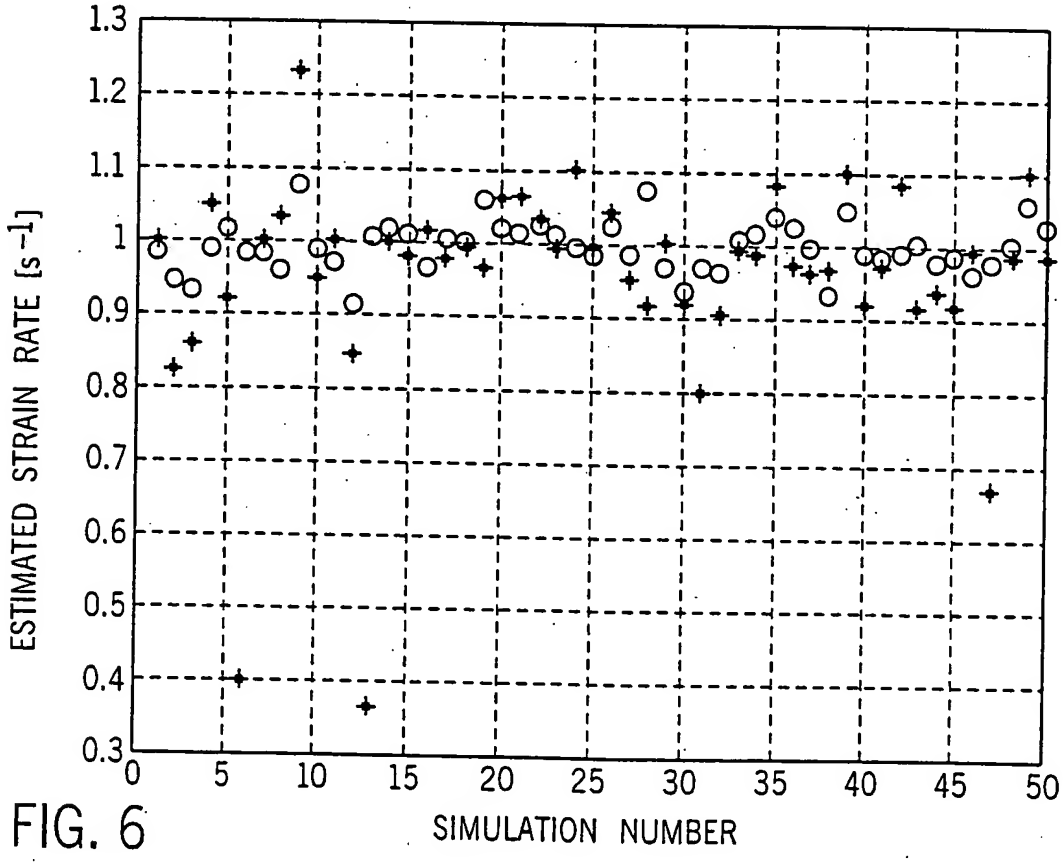

1.3
1.2
1.1
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
ESTIMATED STRAIN RATE [s^{-1}]
0
5
10
15
20
25
30
35
40
45
50
SIMULATION NUMBER

FIG. 6

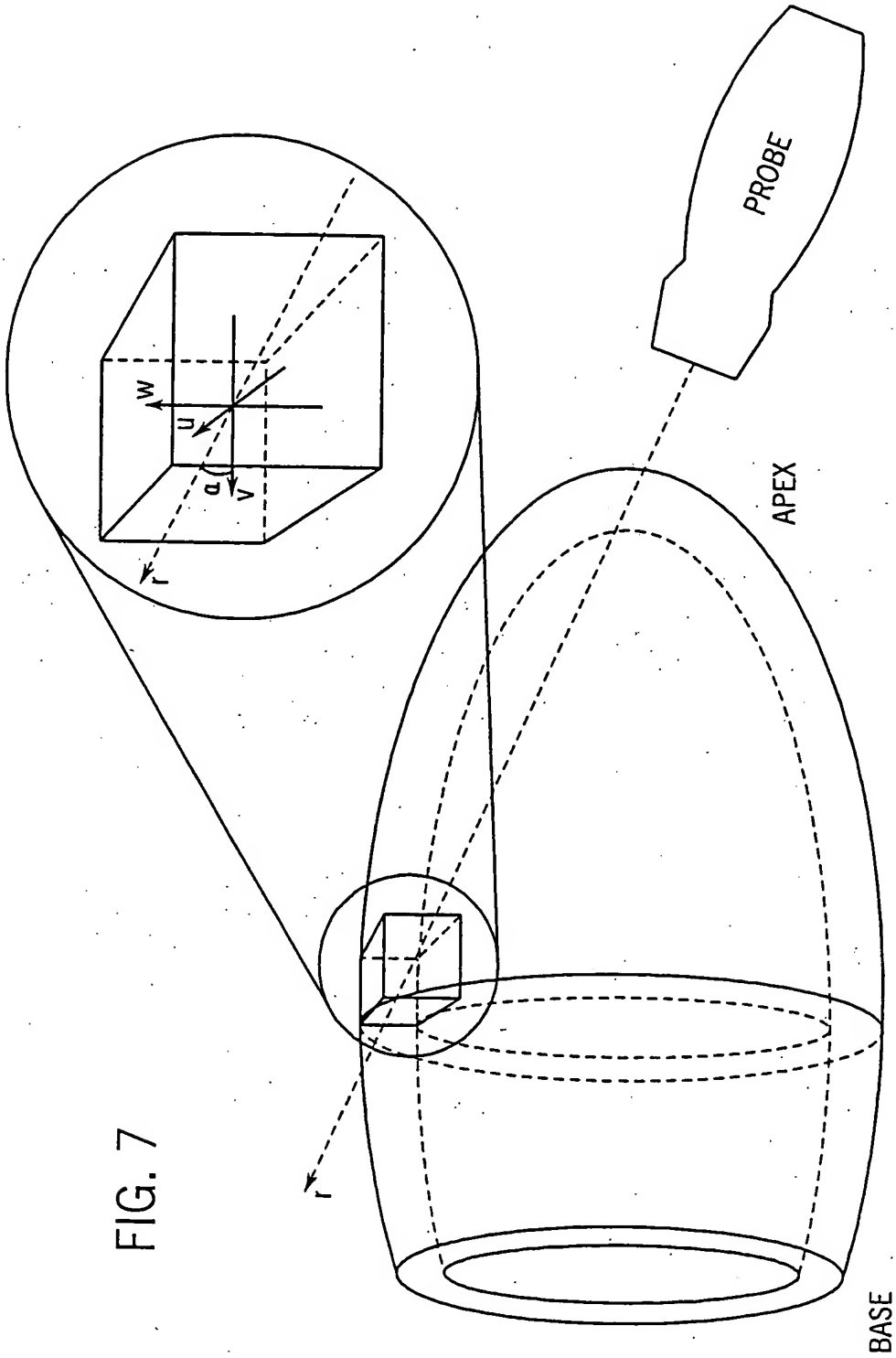


FIG. 7

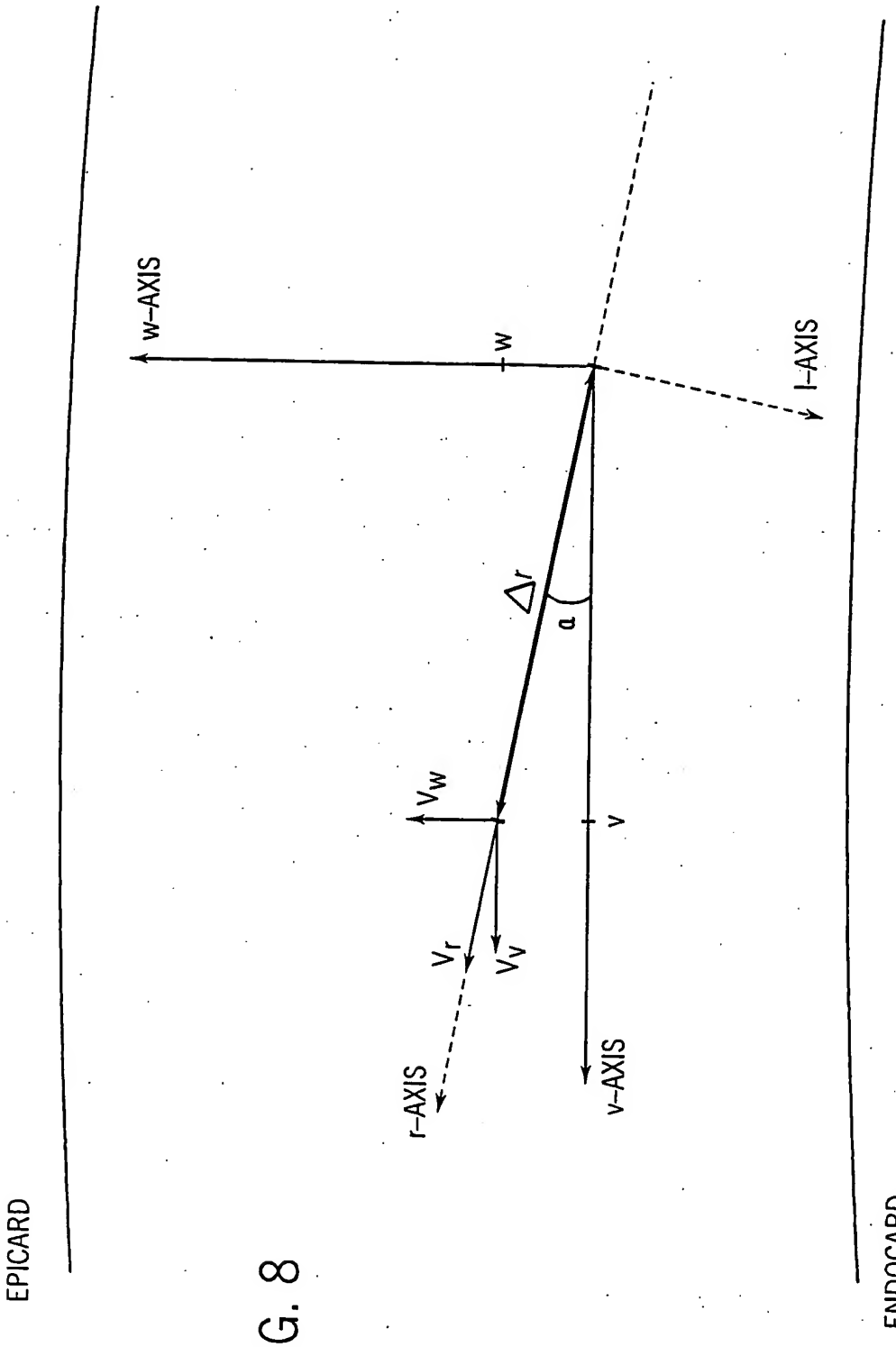


FIG. 8

FIG. 9

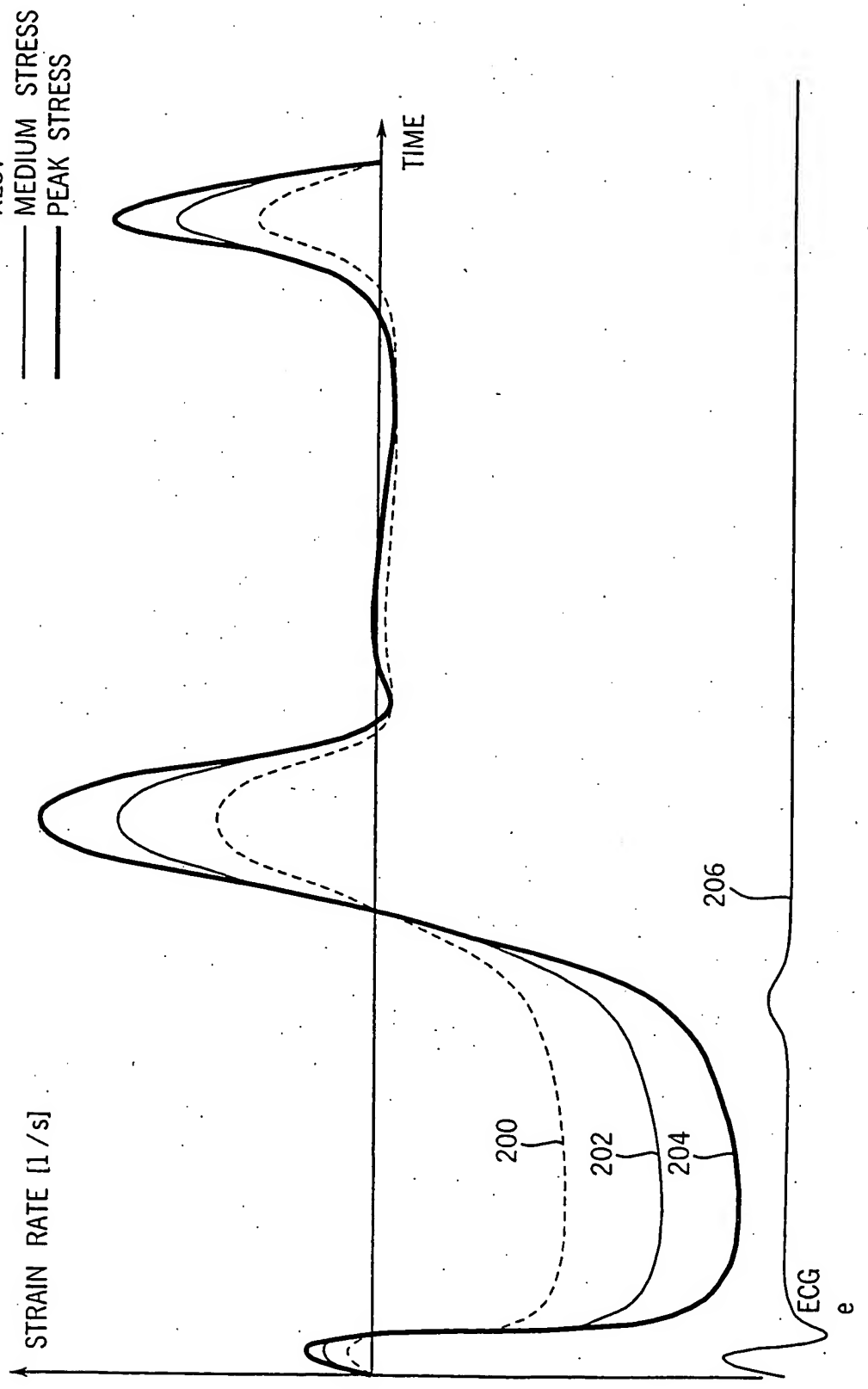

REST
MEDIUM STRESS
PEAK STRESS
TIME
STRAIN RATE [1/s]
200
202
204
206
ECG
e

FIG. 10

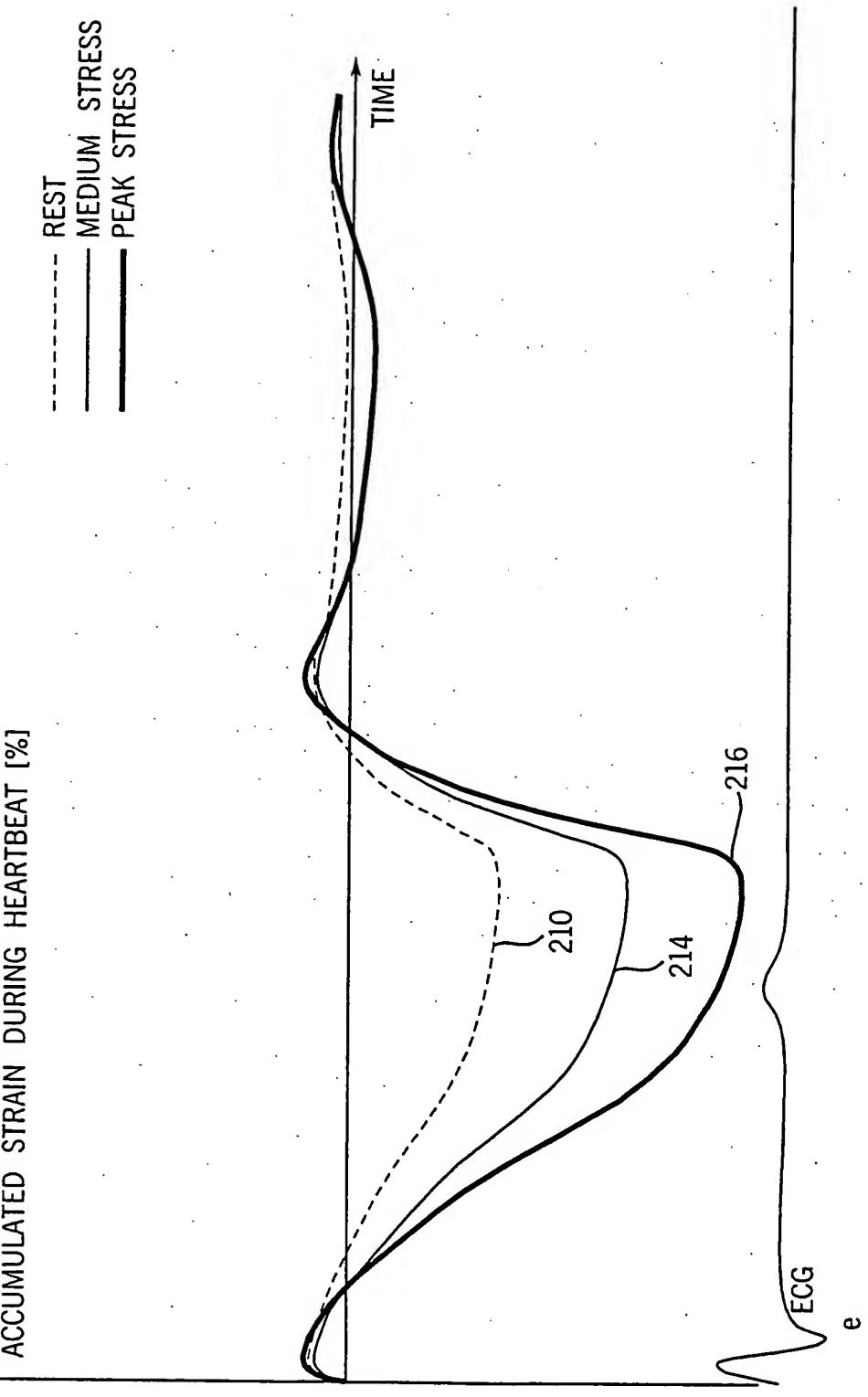

REST
MEDIUM STRESS
PEAK STRESS
ACCUMULATED STRAIN DURING HEARTBEAT [%]
TIME
210
214
216
ECG
e

FIG. 11

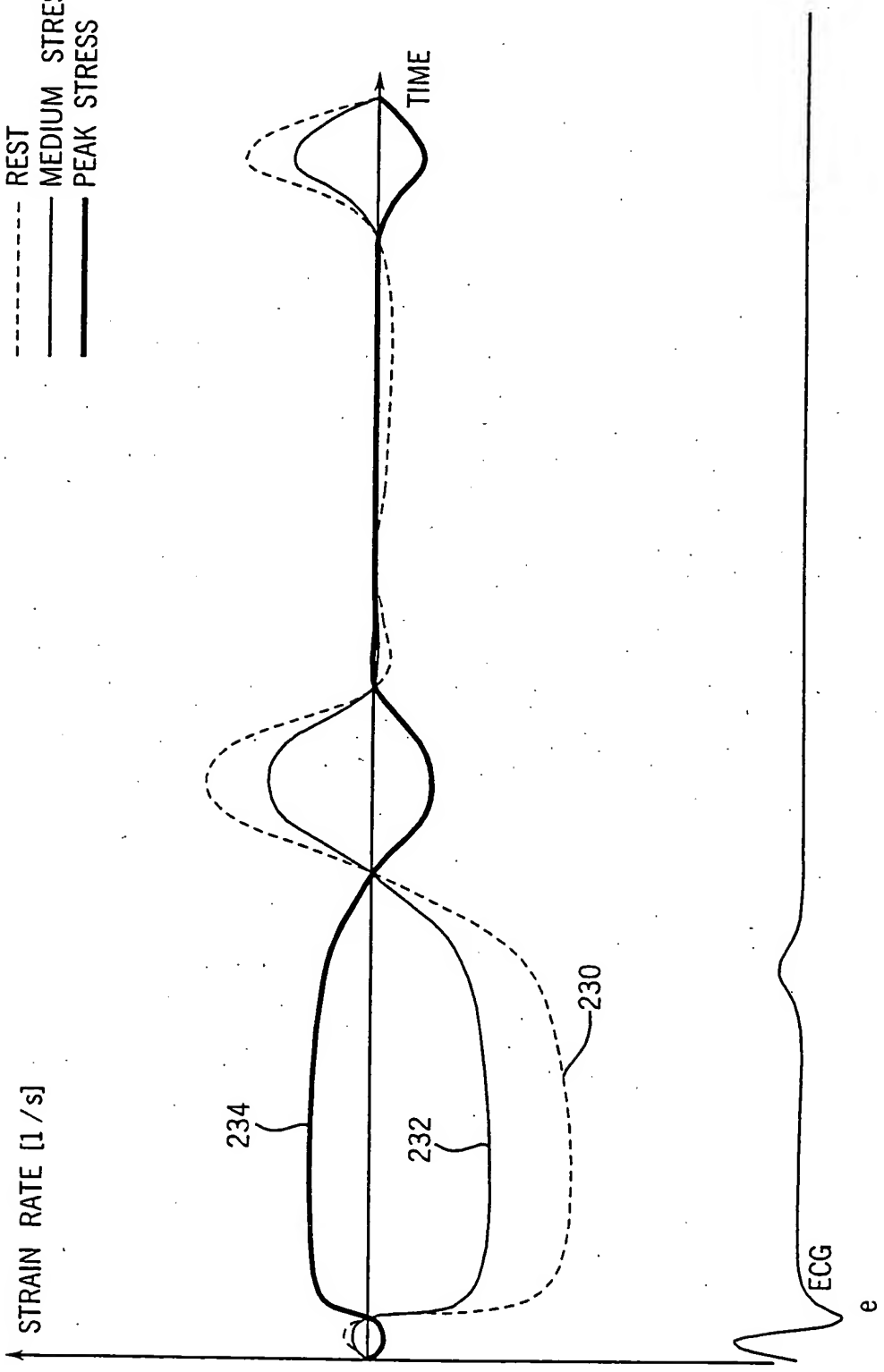
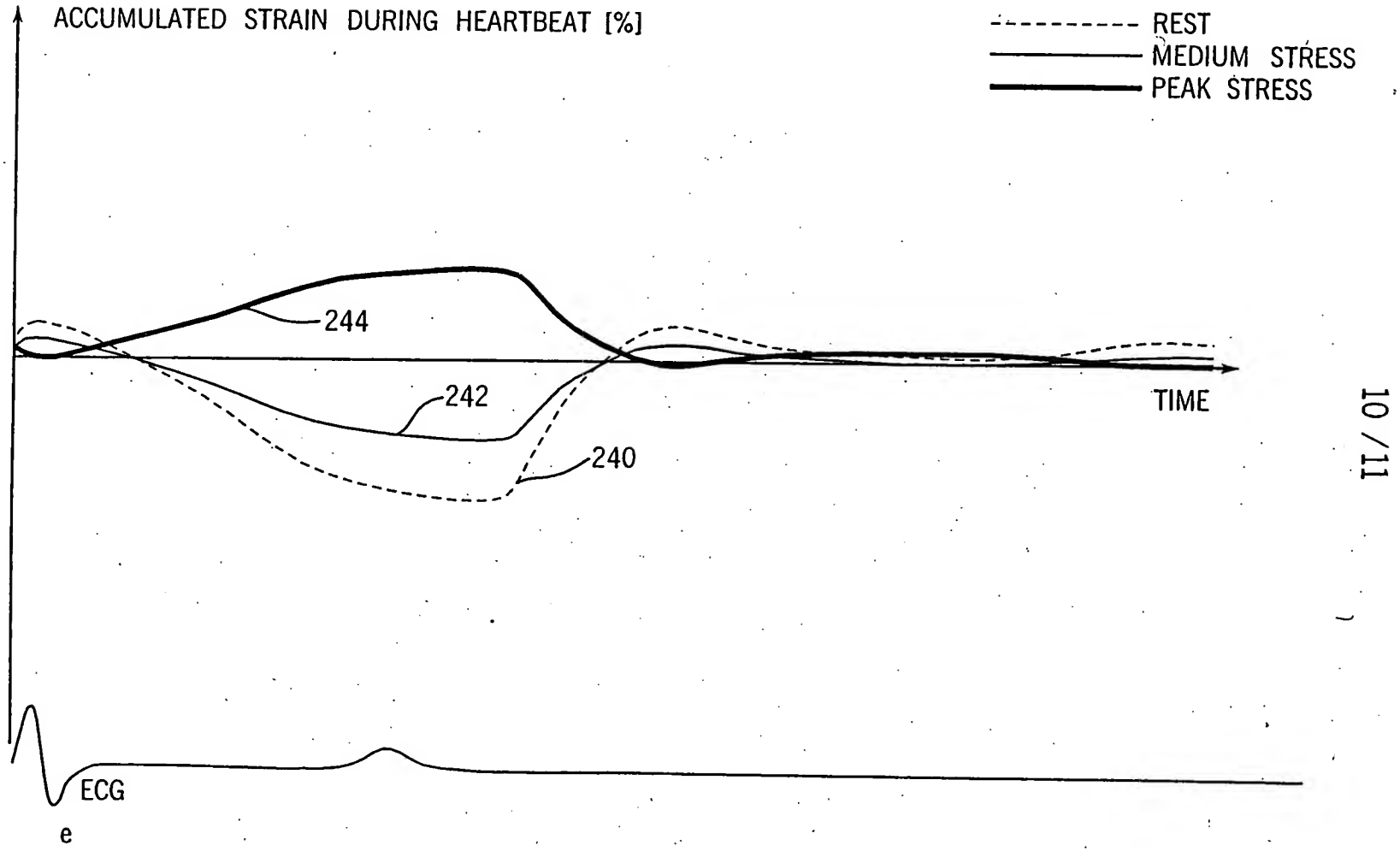

REST
MEDIUM STRESS
PEAK STRESS
STRAIN RATE [1/s]
TIME
234
232
230
ECG
e

FIG. 12


ACCUMULATED STRAIN DURING HEARTBEAT [%]
REST
MEDIUM STRESS
PEAK STRESS
244
242
240
TIME
ECG
e

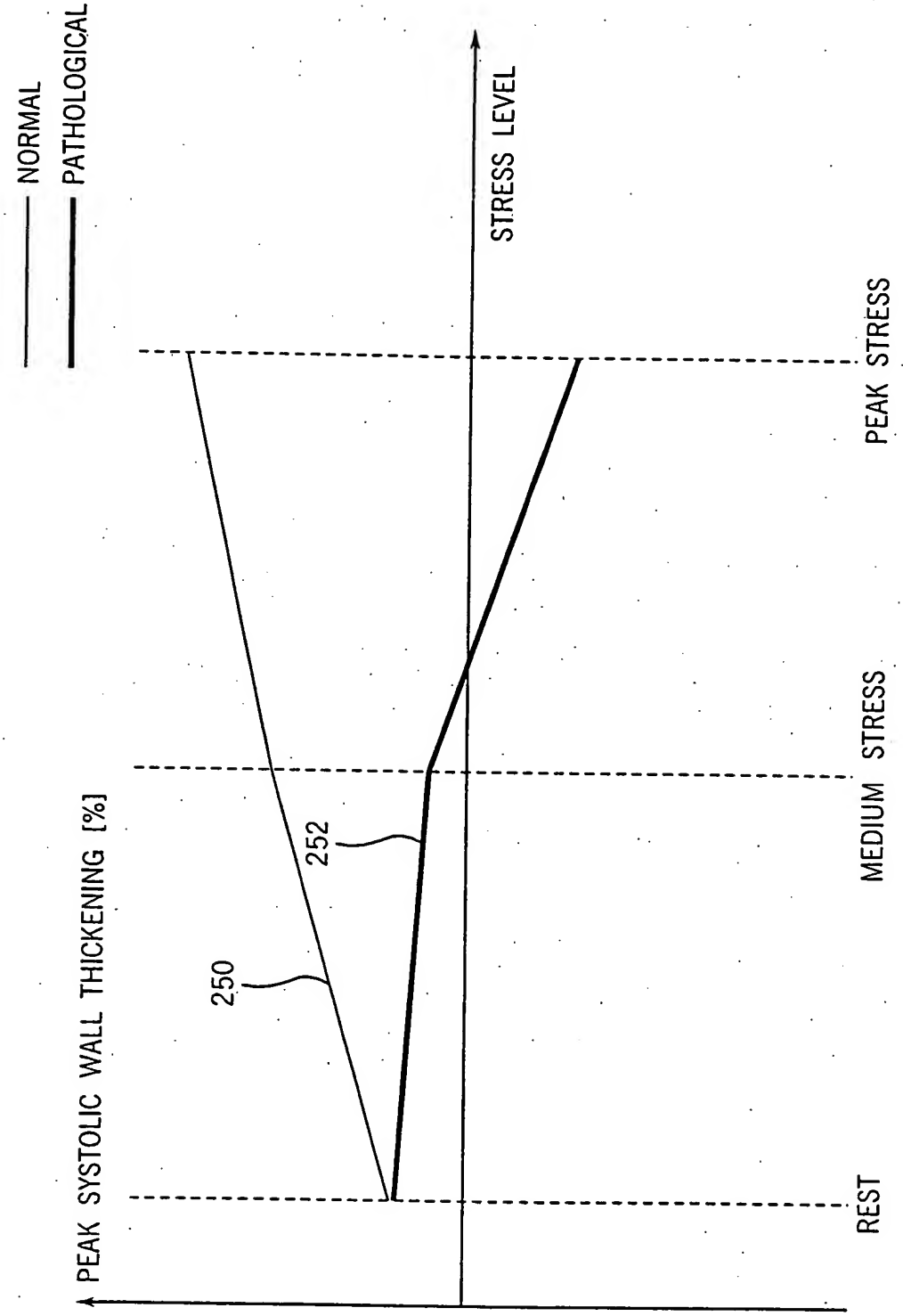

NORMAL
PATHOLOGICAL
STRESS LEVEL
PEAK STRESS
MEDIUM STRESS
REST
PEAK SYSTOLIC WALL THICKENING [%]
250
252

FIG. 13